

**WEARABLE SENSING MODALITIES FOR A NECK-BASED
HEALTH MONITORING SYSTEM**

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WEARABLE SENSING MODALITIES FOR A NECK-BASED HEALTH MONITORING SYSTEM

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SUMMARY

Advances in modern medicine have led to a decline of deaths among youth populations by introducing immunizations and effective treatments for infectious diseases which had previously resulted in high mortality rates. This decline has led to an increasingly aging population, which is estimated to be 20% of the population by 2030. This has led more chronic conditions in the population accounting for 75% of healthcare costs. One method of mitigating this medical burden is the use of personalized eHealth through wearable devices. In this work, a novel multimodal neck-worn device is presented for use in non-invasive health monitoring. A design of a platform for detection of vital signs accessible from the neck including digestive, circulatory respiratory and activity tracking is shown and preliminary results from data collection of wireless tracheal activity using multiple signal modalities is explored.

CHAPTER 1

INTRODUCTION

1.1 Justification for Advancements in Healthcare Technology

Advances in modern medicine have led to a decline of deaths among youth populations by introducing immunizations and effective treatments for infectious diseases that had previously resulted in high mortality rates. As a result of these advancements, society is becoming increasingly older. The National Institute on Aging has estimated that in five years' time the population over 65 years of age will outnumber the population under the age of 5. This figure is expected to continue with an increase in people over the age of 65 by 250% in underdeveloped countries and 71% in developed countries between the years of 2010 and 2050, while the population over 85 is expected to increase over three-fold [1]. In the United States, this rate of increase will lead to the elderly population (65+) becoming 20% of the population by year 2030 according to the Centers for Disease Control [2].

Largely as a result of an aging population and lower mortality from infectious diseases, chronic conditions now account for over 75% of the United States' healthcare costs and are expected to increase by 25% in the next 15 years [2], [3]. Non-infectious diseases are now the leading cause of death, with cancer and heart disease contributing to nearly 50% of deaths each year [2]. While the reduction of pediatric deaths and the prevention of infectious diseases are together a great achievement, they present a new burden on the healthcare system. Currently one in ten Americans now has a chronic

condition, and over 25% have multiple chronic conditions [4]. These diseases include hypertension, cancer, diabetes, stroke, coronary disease, kidney disease, and obesity among others. Associated complications often lead to reduced independence and in some cases disabilities that require frequent care from a care giver. Health care is further complicated by cases of dementia in the elderly in which patients are frequently unable to adequately follow healthcare regimens as prescribed and require additional assistance.

Chronic conditions contribute a larger burden due to issues often requiring indefinite treatment and frequent hospitalization. In some cases, complications of these diseases can be mitigated with corrective measurements, which could result in fewer hospitalizations and thusly, lower healthcare costs. There is a great push towards preemptive medicine in order to identify and diagnose medical issues before they become chronic health problems. Health monitoring is not only effective for elderly populations, but also for those in high-risk situations where constant surveillance of health is needed. These populations include, but are not limited to: children with chronic diseases who often do not fully understand the risks of their behavior and its effects on their health, teens who desire the freedom of independence while having chronic illnesses, and busy adults who may have difficulty in keeping with a health regimen during their daily activities.

In the United States, the Affordable Care Act has further catalyzed the need for personal intervention in healthcare with the aid of medical technology to both lower healthcare costs and allow limited health personnel to take care of more patients. In order to address these growing needs, there is a push for advancements in eHealth. eHealth is generally defined as the use of technology to allow healthcare information sharing and

encompasses many aspects including information exchange between doctors and patients and also the collection of health data [5]. eHealth and its synonyms, telemedicine and tele-health, focus on making the stream of information between healthcare providers more efficient in order to enable more precise treatment. This personalized version of eHealth is coined personal eHealth (pHealth), which is aimed at providing prevention of disease through personalized healthcare [6]. The use of these technologies represents a paradigm shift for the healthcare industry and moves from a reactive industry to a proactive one that can spot and reduce the risk of problems before a pathology becomes chronic and costly.

1.2 Potential Applications of pHealth

pHealth has great potential in aiding healthcare workers and is poised to be the next revolution in the healthcare industry. pHealth promises to help alleviate healthcare burden as it allows more information to be shared between the caregiver and patient. Technology that provides personalized healthcare can allow the caregiver to make more informed decisions about the patients' current status and future treatment. pHealth applications may even provide more insight than the patient is able to give, due to collection of data not dependent on the memory or bias of the patient.

There are many populations that stand to benefit from more personalized healthcare, especially those in constant need of medical supervision, and patients who live in areas where healthcare is not readily available. Potential exists in health tracking for the elderly, in which case data could be presented that shows the complete activities of an elderly patient. Data collected from tracking may indicate the patient's mobility or

gauge how active the person is and could gauge the effectiveness of care for certain conditions such as pain management in arthritis.

There has recently been a push towards pHealth with the development of technologies that enable monitoring of a user's health information such as activity or vital signs. Wearable systems have become available that may provide continuous or semi-continuous physiological measurements of the wearer. An analysis of wearable device sensing is provided in Chapter 2.

Compliance with a medical regimen is an especially important area in which pHealth has potential benefits. In elderly populations, medical regimens can be complicated due to the combination of diseases and disorders that may be present in a patient. One study showed that many peoples in this population take more than one medication daily and have a changing medication schedule [7]. Additionally, intake of food and water is important with those populations suffering from dementia. pHealth devices that provide gentle reminders and track compliance may be beneficial in directing a medical routine. Similarly in adolescents, medication needs and other healthful activities may be poorly understood which requires strict supervision. pHealth devices can be instrumental for supervised tracking in populations where medical intervention is required.

1.3 Wearable Devices

Wearable devices provide a continuous stream of data from a user that may give a much more detailed picture of the health profiles of an individual. The potential of data from wearable devices is large and grows as more sensing modalities are added to the device(s). From collected data, more complete profiles can be used by a doctor or

caregiver to provide insight that can be used to identify potential problems, collect diagnostic data on existing conditions, and eliminate potential issues or problematic recall of the wearer. Many wearable devices can also be made relatively unobtrusive, which gives the wearer the ability to perform activities of daily living while donning the device without any interference. Wearable device technology has been made possible in recent years by miniaturization of microcontrollers, MEMS sensor designs, ubiquity of near-field computers (in the form of smartphones), and the introduction of low power wireless communication. Previously, sensing of body signals required sensor to be wired to locations on the body with an external processing unit. With new technology some signal processing can be performed on the device and further processing can be offloaded to a smart phone via Bluetooth, or other wireless communication protocol. These wearable sensors hold the promise of sensing and preventing illness by providing greater information and capturing episodic data that may not be apparent in the clinical setting.

E-textiles have allowed sensors to be combined into clothing thereby mitigating or eliminating the potential drawback of added additional articles to the body. This form factor aims to mimic the look and feel of normal clothing creating an Other devices have been made in the form of wrist bands and other jewelry-like form factors for easy application and removal.

The MIT MITHril was an early pioneer in the field of wearable health sensors, creating a body network in a vest that tracked vital signs and was interfaced to a PDA for data collection in 2003 [8]. Another well cited system, CodeBlue from Harvard, uses wireless distributed sensors over the body implementing ECG, inertial sensing, and EMG [9]. Recently, research in wearable health monitors have has become increasingly active

with specific healthcare monitoring aims. Soliband, a solar powered wrist worn pulse oximeter, has come out of the NSF Assist group, which focuses on development of self-powered sensors [10]. For treatment of obesity, groups from University of California Los Angeles and University of Alabama have developed systems which aim to monitor food intake from the neck and jaw motion respectively [11]. Other systems have also been developed for monitoring of specific diseases such as Parkinson's, or sleep apnea and can be used in rehabilitative settings [12]. A listing of very recent research in wearable health monitors is available in Appendix B.

Commercial activity trackers have also become mainstream. Activity trackers started with simple inertial sensing provided by accelerometers; however, they have now evolved to include much more advanced sensing modalities such as heart rate, gyroscope and digital compass modules, and sensors for perspiration. Sensor fusion has led to greater advances the applications of wearable technology, from simple step counters to highly accurate workout and fitness trackers. These trackers are now able to assess calorie burn and relative intensity in activity from these extra modalities. Major companies including Samsung, Apple, LG, Microsoft and others have taken notice of the potential of this market segment and have all introduced their own devices. Many consumer devices come in the form of a wrist based solution mimicking bracelets and watches, which have been staples in wearable technology for decades. Other devices have been specifically designed for use by medical professionals or researchers in a clinical setting. One example is the Shimmer platform which is a popular modular sensing platform for wearable studies [13]. An overview of commercial devices is shown in Appendix A.

Wearable devices are able to sense the body by strategically placing sensors to take advantage of physiological signs. Descriptions of various sensing methods are described in Chapter 2. The wrist has been commonly used due to familiarity with user because of another wearable technology, the watch. Other locations such as chest have been chosen for their proximity for sensing. For example, the foot and head are commonly chosen for gait and head impact monitoring respectively. Another location is ear, which has been explored for non-invasive heart rate monitoring and food intake monitoring. One relatively unexplored location is that of the neck. It is thus the goal of this thesis to explore the potential unique sensing location the neck provides to several key physiological signs including digestive, circularity, respiratory and a strategic placement for inertial sensing on the central longitudinal axis of the body.

CHAPTER 2

CURRENT STATE OF WEARABLE HEALTH MONITORING

2.1 Modalities of Sensing

Much research pertaining to measurements provided by wearable systems has been explored. Sensing modalities differ on their function depending on the objective of the use cases. It is useful to have an understanding of the different sensing modalities and their common uses cases in order to develop an idea about how they can be integrated for accurate monitoring. The following chapter gives an overview of sensing modalities used by wearable systems and their uses as sensors in wearable systems.

2.1.1 Optical Sensing/ PPG

Optical sensing in wearable applications typically relies on the technology of light emitting diodes (LEDs) for light transmission and photodiodes or phototransistors to transduce light signal into electrical signal. Optical sensing methods in wearable sensors are most commonly used to extract some information from the body tissue by using light absorption, which measure the difference between transmitted light and detected light allowing for information to be determined about the tissue from a calculated transfer function.

Optical detection is an accurate method of retrieving heart-rate in small wrist-worn devices through the application of optical pulse plethysmography (PPG). PPG measures the relative changes in blood volume from pumping of the heart of by measuring the light absorbance in skin. Typically, PPG has relied on the use of red light

LED's operating at 645-660nm and near infrared (NIR) wavelengths as these provide deep tissue penetration and can detect blood volume change in subcutaneous tissue [14]. Green LEDs operating at a wavelength of 530nm have also been used due to lower presence of artifacts. However, this shorter wavelength has a lower penetration depth and is mostly dependent on fractional blood volume in the dermis layers of the skin [14][15].

Pulse Oximetry is a special application of PPG that can indicate the presence of oxygen in the blood. The mechanism of pulse oximetry relies on the species of hemoglobin consisting of an oxygenated type and deoxygenated type. The detection of the relative volumes of these types in the blood requires the use of two LEDs used in tandem and a difference to be calculated between them. The wavelength of these LEDs is based on the absorbance of certain wavelengths of light of the different species. The red wavelength at 660nm is most commonly used as its absorbance significantly drops with increased oxygen saturation [16]. The infrared wavelength of 940-990nm is then used as a comparison as its absorbance increases with the oxygen saturation [16]. Using the returned signals of light, total oxygen saturation is calculated by dividing the ac component of the signal caused by the pulsatile flow of blood by the dc signal, which is time invariant and is caused by scattering in the tissue with little blood content.

Two main methods exist for detection of blood content through light: transmissive and reflective. Transmissive is the simpler of the two designs as it passes light through the tissue to a detector located on the opposite side of the tissue. This is not an ideal measurement technique due to limitations on positions on the body where the tissue width is thin, such as the finger, earlobe or nose. Reflectance mode PPG provides a solution to this problem as it uses an LED and photodetector located opposite to each

other on the same plane and measures the light reflected back from the tissue rather than the transmitted light. Reflectance generally sensors require more accurate calibration and are more susceptible to noise but operate on the same principle as transmissive types.

Figure 1 shows the configurations for the two types of PPG. The received signal may be improved by using more than one LED arranged around a photodetector and can provide greater accuracy in signal acquisition [17][18].

The use of optical sensors is already prevalent in current wearable sensors. Many fitness monitors use PPG sensor to determine heart rate for activity monitoring. There are many applications of using pulse oximetry to measure the oxygen content of the blood, including circulation information, respiratory distress from lack of oxygen, and breathing rate [19]. These devices are undergoing active research for lowering power requirements. SoliBand project, an NSF assist project, which aims to produce a solar powered PPG on the wrist [10].

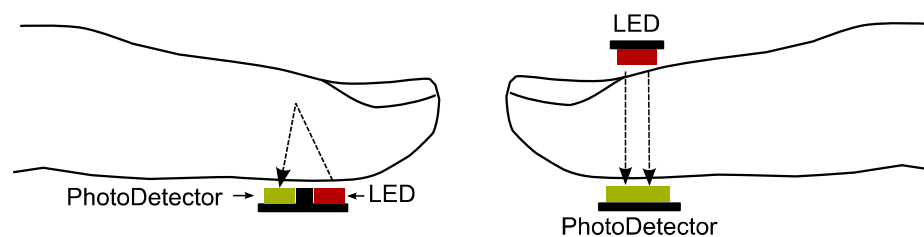


Figure 1. Reflective and transmissive PPG depictions.

2.1.2 BioPotential Sensors

Biopotential signals represent a class of signals created by voltages generated in the body. These signals are generated from the nervous system and propagate to other excitable tissue. This propagation can be detected with the use of electrodes attached to the body. The electrically excitable tissues in the body consist of nerves, and muscle. Excitable tissues are tissues, which react to an action potential. The action potential is the propagating electrical signal in the body, driven by relative ionic concentrations of the cells. Action potentials are characterized by a depolarization caused by a charge transfer from a source tissue, followed by a repolarization and refractory period. Figure 2 below depicts this action.

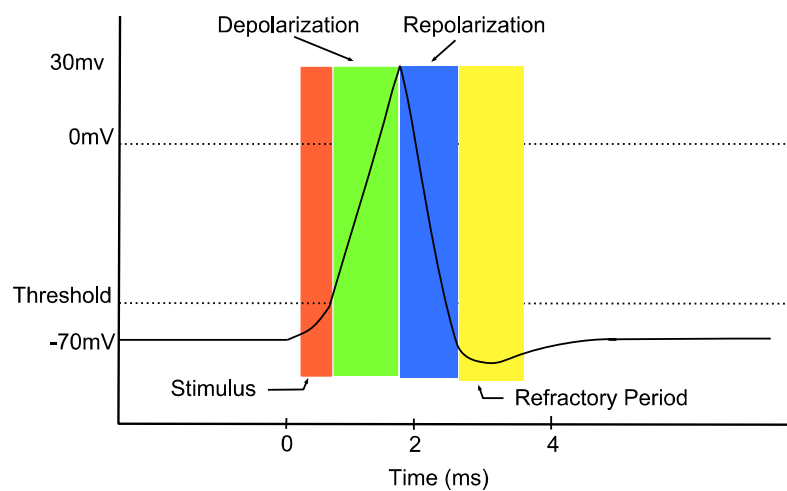


Figure 2. Action Potential waveform depicting ideal waveform from excitable tissue.

There are several common sensing applications which rely on this signaling feature of the body including electroencephalogram (EEG) - measurement of neural activity in the brain, electromyogram (EMG) - measure of muscle activation, and

electrocardiogram (ECG) - measure of cardiac activity. Though the underlying principles of sensing are very similar for these applications, each has specific features and processing to produce information about their respective systems.

The acquisition of electrical signal from the body depends first on electrodes to detect signals. In wearable systems, these electrodes are applied to the surface of the body to detect signals propagating through the skin. The most common electrode is the silver chloride electrode, which provides excellent readings of biological processes that are the result of ion current [20]. For best contact, electrodes may be coupled with an electrolytic gel. These gel electrodes are often coupled with an adhesive in order to provide stable direct contact with the body. Due to discomfort of the wet electrodes, dry electrodes have been implemented which do not require a wet connection, but are generally more susceptible to artifacts and do not provide as low an impedance connection as the wet variety due to a more capacitive interface. More recently, non-contact electrodes have been introduced which provide more comfortable usage during the wearing of the EMG device due to the advantage of not requiring direct application to the skin [21]. Conductive thread electrodes have been also been used in e-textile applications but are subject to changing properties due to absorption of moisture or stretching in the surrounding fabric [20].

2.1.2.1 EMG

Electromyography (EMG) is the measure of electrical impulses caused by the activation of muscles in the body. The signal is a pulse train signal with frequency determining the strength of the muscle contraction. These signals can be retrieved using electrodes on the surface of the skin in the application of surface EMG (sEMG). sEMG

provides a superposition of the signal coming from the muscle groups in the vicinity of the electrode. EMG in this paper will refer to the non-invasive sEMG method. Larger muscle groups with stronger action tend to give larger signal.

Processing of EMG typically follows a standard algorithm where a band-pass signal is applied to remove any noise that is not in the frequency range of the expected EMG signal. After this filtering, the envelope of the signal is taken to provide an idea about the activation from the amplitude of this result with higher amplitude corresponding to a stronger muscle action.

Surface EMG has been used in several applications in wearable health measurements where muscle activation can be indicative of other physiological signs. Uses of EMG in wearable sensors include posture control, gait, activity measurement, epileptic seizure detection, and of particular interest, swallow detection.

2.1.2.2 ECG

Electrocardiogram (ECG or EKG) is the measure of the electrical signal of the heart. This signal is caused by the neural activity of the heart that causes the contraction in a heartbeat. ECG signal follows a common waveform pattern in health people between each user with only small variations. The waveform consists of distinct parts, which have physiological interpretations of cardiac activity. Figure 3 shows an ECG reading of a healthy cardiac cycle consisting of the P-wave, the PQ segment, QRS complex, ST segment, T wave and the U wave. It should be noted that this waveform is a representation of the superposition of all signals of the heart and is the result of several propagating waves. For simplification, only the entire waveform will be focused on here. The first section of the ECG is the P wave which is the signal contributed by the

electrical impulse generated at the Sinus node and the activation of the atrial muscle as well as conduction in the AV node. The PQ segment is signal generated by the propagation of the electrical signal along the Purkinje fibers from the AV node to the muscle wall of the ventricles. The large spike or QRS complex in the ECG is representative of the depolarization of the heart muscle which causes contraction of the ventricles. The T wave is caused by repolarization of the ventricles and the ST segment is the refractory period between the depolarization and repolarization of the ventricles. The timing and amplitudes of these signals can be used to determine pathologies of the heart either by automatic detection algorithms or the transfer of data to a trained physician.

Wearable ECG sensors have been developed and demonstrated in chest patches as well as other areas for determining heart rate [22]. An R-wave signal can be picked up at many locations on the body allowing for heart rate calculation in multiple areas. Heart rate sensing has been used to help define activity status of the wearer and when coupled with inertial sensing has been used to provide an estimate of caloric expenditure.

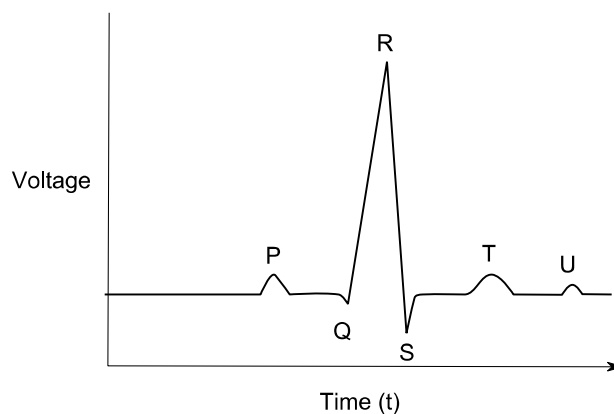


Figure 3. Representative healthy ECG waveform depicting the P,Q,R,S,T and U wave portions of heart electrophysiology.

2.1.2.3 EEG

Electroencephalography records the electrical activity of the scalp in order to read signals being passed by the nerves in the brain. EEG typically uses an array of electrodes spaced around the head to determine signal propagating from different areas of the brain. In wearable systems, dry electrodes are used for comfort, requiring no gel to be applied to be applied to the scalp. For data analysis, potentials recorded at each electrode are combined into a vector field and statistical analysis is applied in order to attempt a measurement of some function.

EEG is potentially very useful in wearable health systems for recording brain activity of related disorders which may present irregular neural activity such as ADHD, PTSD, mental fatigue and emotional responses [23] [24][25]. EEG signals indicating certain mental disorders may also be used for diagnosis or tracking.

2.1.3 BioImpedance

Bioimpedance refers to the electrical properties of the body. This sensing method measures the impedance of various tissue structures and is based on dynamic physiological processes. Changes in the impedance can lead to determination of changes occurring in the tissue. The collection of bioimpedance is based on the same principle as network analysis in electrical engineering, and is performed by injecting a non-harmful AC current into the body. The return current is then analyzed to form an impedance measurement based on the changes to the signal [26]. Frequencies are specifically selected to detect different processes as the response varies between different biological tissues [27]. There are two basic implementations of bioimpedance: 2-electrode and 4-electrode. The former uses the same electrodes to transmit and receive the signal while

the 4-electrode system uses dedicated transmitters and receivers. The 2- electrode setup suffers from larger artifacts and a lower frequency range for which a response can be measured.

Bioimpedance markers can be used to determine fluid composition of the body [26]. This is an especially useful marker in chronic conditions such as heart disease or diabetes where fluid retention may be an indicator of the progression of the disease [28]. This sensing system has also been used to determine overall body composition, which provides an estimation of body fat, muscle mass and bone mass. This application is especially helpful in tracking obese populations to determine how body composition is changing in a certain patient population and is especially useful when combined with other activity sensing modalities to provide an overall profile of health. Heart rate measurement has also been implemented using bioimpedance to extract fluid flow in tissue. [29], [30].

2.1.4 Galvanic Skin Sensor

Galvanic skin response (GSR) measures the electrical conductance of the skin in order to determine moisture and ion content. GSR operates by production of a non-harmful DC voltage through a skin contact electrode and measures the resistance by recording the voltage that is potentiated on a second electrode in contact with the skin in close proximity. This is similar to the measure of bioimpedance except that the process is simplified by only considering conductance of the skin and operating with DC sources.

Galvanic skin responses may be used to determine disturbances in the autonomic nervous system which allows the sensors to detect emotional responses such as stress which can reveal itself through perspiration in certain areas of the body[31][32]. Poh et

al showed a measured increase in skin conductance during seizures using a continuous GSR monitor indicating that GSR may be used in the detection of seizures [33]. These sensors may also be used to detect perspiration caused by activity and have been used in the determination of the intensity of an exercise, they can also measure a marker for diabetes in dynamic sweat tests where neural effects from diabetic neuropathy limit sweat volume [34].

2.1.4 Acoustic Sensor

Acoustic signals are the signals acquired by the use of a microphone or other vibrational sensor. A microphone in this case is defined as a device that is usually used for detection of mechanical vibrations through the audible sound range (20Hz – 20KHz) though it may also detect signals outside that range. There are several types of microphones used in wearable sensors supporting the recording of these signals including electret, MEMS, and piezoelectric [35]. Electret condenser microphones are the most popular microphones due to their low cost and simple implementation [36]. Piezoelectric microphones are used in some contact situations as they can be directly coupled to a surface without an air-gap for the detection of sound vibrations in the material [37]. MEMs microphones provide a very small packaging size with good performance. The MEMs microphones operate using several principles including piezoelectric where voltage is produced by strain on a microscopic membrane. Capacitive sensing may also be implemented where deflection of a capacitor is measured and converted to audio signal.

Acoustic sensors have been applied to wearable sensing systems which mimic traditional healthcare techniques of monitoring respiration and heart sounds [38]. With

continuous signals made available, wearable acoustic sensors have also taken on other applications. Acoustic signals may be able to detect and classify the occurrence of certain signals that may indicate sickness (coughing, sneezing shortness of breath); they may also be able to extract data from audio signals such as laughter, deep breaths and yawns [37][39]. One system built in association with Microsoft dubbed the BodyBeat has explored tracking certain common activities such as showering, eating or brushing of teeth with an acoustic monitoring system [39].

Of particular relation to this work is the use of acoustic sensing to track tracheal activity such as swallowing or chewing, which requires processing and classification of the acoustic data emanating from the throat [40]–[42]. This type of monitoring has implications for tracking food ingestion, water intake and medical compliance.

Ultrasonic sensors, though not in the acoustical range of frequencies, operate using very similar technology. In this case, piezoelectric devices are used to transmit data for reflection off of biological structures for analysis. The collection of the ultrasound data follows the same principles of the acoustic signal detection by using piezoelectric sensors to detect the reflected vibrations. These sensors can be used to detect heart rate or respiratory rate [43].

2.1.5 Temperature Sensor

Temperature sensors are used to detect skin temperature in wearable systems. There are many technologies used in the application of temperature sensing including: optical fiber, infrared (IR) detection, resistive (thermistor), thermocouples, and silicon implementations [44]–[48]. Besides IR, each of these listed require direct thermal coupling to the skin in order to detect the temperature. IR has an advantage in that it may

detect the skins temperature without this direct coupling and can resolve the temperature of the skin quickly without a lag time that may be caused due to thermal load.

Skin temperature may be used to determine basic physiological symptoms such as fever or overheating and may also be used in conjunction with other activity monitoring sensors to determine the energy expenditure or relative intensity of an activity [47], [49].

2.1.6 Force and Stress Sensors

Force sensors are a class of sensors that are used to detect forces produced on the sensing body. The most common force sensors are piezoelectric sensors in wearable sensors. Piezoelectric sensors transduce mechanical stress to electrical signal through the piezoelectric effect. The piezoelectric effect is a phenomenon that results in electric potential energy from a stress on a crystalline structure. Piezoelectric force sensors in the wearable context can be useful in detection of flex and pressure applied to the vicinity of the sensor. Along with piezoelectric resistive strain gauges may also be used which change resistance based on the amount of stress perceived by the structure. This change in resistance can then be monitored using a wheatstone bridge circuit.

There are many potential uses for force sensors in wearable systems, though most are used to detect biomechanics of movement such as angle changes of joints, and chewing [50]. The flex sensor may also be useful in detection of swallowing from throat deformation that occurs in the swallowing process.

2.1.7 Inertial Sensors

Inertial sensors are a grouping of sensors that collect motion data from the subject. The main two sensors comprising this category are accelerometers and gyroscopes. Accelerometers give information about linear acceleration and are useful in

determining position and movement of the subject relative to the Earth's gravitational field, which always produces an acceleration component. Gyroscopes measure the angular changes in velocity about the axis orientation of the sensor. Advancement in microelectromechanical systems (MEMS) technology has allowed these sensors to be produced in small package sizes with low power requirements and made them an ideal sensing modality for wearable sensors

Along with accelerometer and gyroscope, magnetometers are also being implemented on the same package in order to create a digital compass, which aims to sense complete orientation in space by obtaining the angle to the Earth's gravitational field and magnetic field, along with movement within this field. Altimeters have also been included in this sensing strategy to provide elevation through the comparison of an internal pressure and the atmospheric pressure surrounding the sensor. With the inclusion of these sensors the orientation and movement data can be determined.

These sensors allow for a variety of applications with the most common being inertial activity tracking such as walking or running. Inertial sensing has also been implemented to detect heart beats by implementing a ballistocardiogram (BCG) through the use of an accelerometer [51]. Wearable seismocardiography has been implemented for measuring heart beats using the vibrations during cardiac activity [52] and respiration rate has also been calculated using an accelerometer located on the chest [53]. Inertial sensors are being used for monitoring of rehabilitation of injury and gait detection and have been explored for automatic fall detection for use in the elderly to provide a real time representation of posture [22], [54]–[56]. Another valuable feature of inertial sensors is their inclusion in multimodal platforms as a method of distinguishing inertial

influence on other sensitive sensors subject to motion artifacts. Using the inertial data, motion artifacts may be identified and compensated for or eliminated.

2.2 Locations of Wearable Devices

The application of sensors and use of the wearable system depends greatly on the locations of the sensors. Many of the sensors listed in the previous section rely on close contact with physiological signals that can only be detected at certain points on the body. Certain areas of interest have risen, including the wrist, chest, feet, upper arm, lower limbs, and head, with the number of devices at the wrist outweighing the devices at other locations. Figure 4 provides an overview of locations of sensors and the types of signals that may be extracted from these locations.

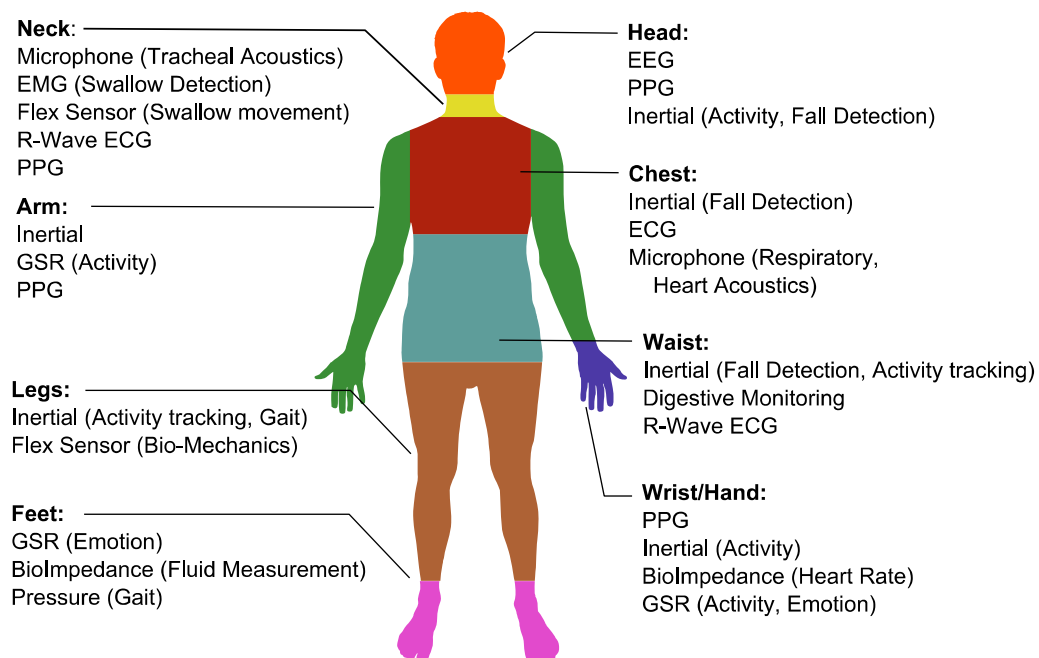


Figure 4. Placements of wearable sensors on the body with description of signals that are viable from each location for use in health monitoring.

2.2.1 Upper Arm

The upper arm offers an accessible placement for wearable sensors without adding bulk to the end of the arm, which allows larger devices to be placed on the arm where they might not be suitable for the wrist. BodyMedia offers a commercial product that detects GSR, inertial activity through an accelerometer, temperature and body positioning for use in medical study. Sazonova et al demonstrated an expenditure prediction device, which used PPG, temperature and an accelerometer based pedometer to achieving a prediction with only 5.77% error from VO2 expenditure readings using all sensors as a multimodal system [57].

2.2.2 Wrist and Hands

The wrist has been a special area of interest largely because of familiarity with other wrist worn devices such as watches. The majority of commercial health tracking devices use the wrist, including those from Jawbone, Fitbit, Microsoft and others. The major drawbacks of the wrist location for wearable systems are a large distance to the core of the body and continuous movement through the limb during daily activities. Movement of the arm requires inertial tracking to implement more complicated algorithms for activity recognition and lowers accuracy in recognition of activities that require a higher accuracy such as fall detection [58].

PPG and Bioimpedance sensors may use the wrist to determine heart rate [17], [59]. Wrist based PPG however has been shown to have a lower accuracy compared to other areas of the body with a large flow of blood closer to the surface of the skin [17]. A collaboration with Texas Instruments to create a BioWatch showed that ECG R-wave is also accessible from electrodes located on the wrist allowing for another heart rate

sensing modality [60]. The use of GSR sensors can also be used on the wrist to determine stress levels and emotional state [61].

Sensors on hands have used a glove or ring type form factor [44]. The hands provide many of the same features but have some added sensing features. The small diameter of the finger allows for transmissive type PPG and pulse oximetry to collect data from the tips of fingers with the use of a PPG probe [62]. The finger has also been shown to have better GSR response to stress levels in the body than that of the wrist [62].

2.2.3 Waist

Sensing systems in the thorax have been implemented as belt harnesses, belt clips and patches that adhere to the thorax. GSR, R-Wave ECG, temperature and inertial activity may all be recorded from the waist [63]. Peng et al demonstrated an inclusive wearable belt system for the collection of ECG data with integrated processing and inertial tracking [64]. Activity tracking at the waist has shown to provide good accuracy for detection positional activities like lying down as well as low complexity activities including walking [58]. Fall detection through the use of accelerometers has also been shown to be more accurate than that of the wrist, in some algorithms providing sensitivity of 97% for fall events [54].

The waist may also be used to identify signals that are indicative of digestive physiological signs by using acoustic recording for use in post-operative care [65]. A system designed by Proteus Digital Health also uses the location around the stomach to receive wireless signals from an RFID during digestion to indicate medication compliance [63].

2.2.4 Chest

The chest provides good positioning for direct reading of physiological signals emanating from the heart and lungs. This location provides good electrical contact for ECG readings. Respiratory signals can also be captured from the chest through inertial sensing of chest rise and falls, acoustic through digital stethoscope applications and ultrasonic applications. Due to the central location on the body's axis, the chest also provides accurate inertial sensing for activity recognition which is similar to performance from activity tracking of the waist [54],[58].

The form-factor of wearable systems focusing on the chest location includes vests, harnesses and adhesive patches [28], [66]–[68]. E-textiles also provide an attractive option for chest sensing applications due to the ability to be integrated into a shirt or vest providing a less obtrusive device.

2.2.5 Lower Limbs

The legs through the thigh and knee provide a good placement for recognition of more complex movement platforms through inertial sensing, though this location does not provide as accurate results for fall detection as the chest, neck or head [58]

The feet provide a strategic placement for detection of gait cycle analysis and activity measurement from walking or running movements. Accelerometers have been used to estimate energy expenditure from the feet position. Gait analysis is performed from this location using inertial and force sensors for detecting step force and leg swing [57], [69]. Piezoelectric pressure sensors have been used to detect gait changes during rehabilitation [70]. Ultrasonic sensors have also been placed into the shoe in order to detect changes in tissue from walking pressure in diabetic patients [71]. In addition to inertial activity and force measurements, GSR sensors may be used at this location to

detect emotional status in a similar manner to the hands as well as indicate neuropathy from diabetes [34]. Healy et al, in a collaboration with Intel Labs, showed the integration of GSR sensors in a sock providing comparable results to other worn GSR sensors for comfortable monitoring of emotional response [72].

2.2.6 Head and Neck

The head and neck provide attractive placement for sensors due to a centralized location of various physiological systems. The neck especially provides close contact to many of the body's systems including circulatory, digestive, respiratory and neural. The neck can provide heart rate sensing through PPG and even blood pressure measurement through the use of MEMS pressure sensors [73]. By virtue of being located on the main axis of the body, the location provides accurate inertial activity monitoring. The neck has been shown to have a higher accuracy in fall detection when compared to wrist, waist and lower limb when using simplified algorithms [54]. Study showed 80% sensitivity and 100% specificity in detection of falls from the neck using a real-time algorithm [74]. The detection of ECG R-waves has also been achieved from the neck location using a non-invasive wearable neck band for the determination of heart rate [75].

The ear provides good placement for wearable monitors with a location that already has widespread adoption of wearable technology following similar form factors to headphones and hearing aids. This placement provides a good mounting location for inertial sensors capturing data at the head. The ear also offers a location able to record transmissive PPG from the earlobe and reflective PPG from the ear canal with faster responses to changing oxygen concentrations than finger sensors [45][76]. He et. al explored behind the ear as a location for both PPG and ballistocardiography for

measurement of oxygen saturation and arteriole pulse transit time with favorable results when compared to clinical ECG readings of the heart [77]. The feasibility of recording EEG from the ear using an in ear electrode set has also been shown, though the applications of ear EEG are mostly unexplored [78].

Swallow and chewing detection are unique to the throat area with the ability to monitor tracheal activity and jaw motion from proximal sensors. Sazonov et al showed a chewing monitoring system which aimed to detect food intake by monitoring the chewing signal captured from a piezoelectric flex sensor on the jaw [50]. Swallow detection has been attempted to be captured from EMG, impedance, acoustic and flex sensor measurements. Impedance pharyngography (IPG) has been attempted as a method of measuring bioimpedance changes in the neck caused by the pharynx during movement. Studies using IPG as a swallow indicator have found that while feasible it is subject to the movement artifacts caused by neck motion and also is greatly affected by differing anatomy of necks between subjects [79], [80]. Amft et al demonstrated a system using submental and infrahyoid throat surface EMG recording and acoustic data captured with a microphone that achieved a 0.73 detection rate for viscous foods [81]. Acoustic interpretations of swallows have also been explored [42]. Olubanjo et al showed a real-time acoustic algorithm achieving a .79 recall rate from a throat microphone. Piezoelectric flex sensors have also been used to detect neck displacement during a swallow [82], though these sensors suffer greatly from other motion occurring at the neck during activities of daily living. Integration of several of these methods may be able to provide a real-time system for monitoring swallowing from a neck-worn system

The neck provides a unique placement on the body for the establishment of a health platform due not only to the unique sensing opportunities but also for user compliance. People have already become accustomed to wearing devices around the neck similar to necklaces. However, exploration into the area of the neck has been mostly performed using only individual sensing modalities. For this reason, the neck will be heavily focused on in this work as a novel location for a sensing platform.

2.3 Wireless Communication for application in Wearable sensors

2.3.1 Bluetooth

A key feature for wearable sensors is the ability to easily transfer the data collected to other systems for cataloging and processing. Wireless data communication pathways provide the ability to easily transfer this data in real-time. Several wireless standards for data communication exist including Wi-Fi, Bluetooth, Zigbee, ANT+ and cellular data networks. Wi-Fi and cellular 3G/4G technology provide high performance protocols for wireless data transfer, however, they have much higher power requirements due to larger transmission line of sites and larger data processing overhead. Due to these limitations, these choices are not well suited for production of very low power wearable systems. Bluetooth, ANT+, and Zigbee provide lower power technologies for applications in body area networks (BANs). Zigbee is a mesh network technology that allows each device to link independently of the other and has been used in several applications to provide a wireless link between sensors in a BAN [49], [83], [84]. For communication with external devices, Bluetooth 4.0 provides the best option because of its inclusion in many consumer electronics including smartphones, tablets and computers for storage and additional processing on the data received. Bluetooth makes use of

various profiles that describe high-level protocols for transmission of different data for different uses (audio, serial data, phone book, file transfer, etc...). These profiles can be leveraged for quick development and interoperability with many devices. ANT+ is a proprietary protocol that has been developed for personal area networks for the linking of devices, however, its use is more limited due to its lower inclusion in consumer electronics. Recently, Bluetooth 4.0 has become available and has introduced Bluetooth Low Energy (BLE), which is a low power protocol aimed for communication between mobile devices. BLE has a lower peak current draw than ANT+ and has a very high efficiency for data transfer [85]. The power specifications and inclusion in consumer devices makes Bluetooth a good protocol communication with wearable sensors.

BLE is well suited for wearable devices and has adopted several health sensing services in its standard, including heart rate, blood pressure, and thermometer profiles [86]. The BLE protocol has a 1Mbps link data throughput and makes use of a client-server model for the transfer of data. In this mode of operation the BLE device is configured as a server hosting the Generic Attribute (GATT) profile service. This service allows requests by the receiver, usually a computer or smart-phone, or the server can send attributes to the receiver in a notification setting. Bluetooth 4.0 specification also includes Enhanced Data Rate (EDR) for faster data rates used for applications such as real time wireless audio transmission.

2.3.2 RFID

Radio frequency identifier (RFID) is a technology that allows unique identification (ID) through the transmission of radio frequencies (RF). RFID is implemented with a reader and a tag. The tag contains the identification number and the

reader extracts this number by emitting an RF pulse. A tag may be either passive with no power source located on the tag, active, which uses a power source to transmit the identifier, or semi-active, which uses power from the receiver for broadcast of data but also contains a voltage supply. In a passive mode tag, power is received from the reader and power is rectified to output a DC component, which powers the internal circuitry to transmit the tag ID back to the receiver. The passive mode provides many benefits as it allows the tag to be very small and cheap. RFID may operate in multiple frequency segments consisting of low frequency (LF) (120-150kHz), high frequency (HF)(13.56MHz), short range ultra-high frequency (433Mhz), North American/European ultra-high frequency (UHF) (902 – 960Mhz / 860-868Mhz), and microwave (2.45 – 5.8GHz) [87].

Near Field communication is a special application of RFID operating in the HF range developed in collaboration between NXP Semiconductors and Sony. NFC communication provides two-way communication as a receiver can act as a tag. NFC also has integrated security measures for use in payment systems. Several NFC protocols have been adopted as ISO standards with the most common being ISO/IEC 14443-A.

In wearable systems RFID has several uses. One is the implementation as a communication protocol for sending data to external devices creating sensor nodes on the body. Yang et al implemented such a device for transmission of temperature data using a conductive ink based RFID antenna design [88]. Another use of RFID systems for wearable health monitoring systems is in medication compliance where the medication is equipped with an RFID tag and detected through a reader worn on the body. Such a system has been developed by Proteus which uses makes use of an active tag system that

receives power from a reaction between chemicals in the stomach in order to send its identifier to a receiver patch worn on the skin [63]. A UCLA group has also shown proposed an UHF (860-960Mhz) RFID pill design for detection from the throat [89].

CHAPTER 3

PRELIMINARY ASSESSMENT OF A NECKWORN SYSTEM FOR THE MONITORING OF TRACHEAL ACTIVITY

As mentioned in chapter 2, the neck is an advantageous location for the collection of health data. In order to develop algorithms for detection of tracheal activities a preliminary data collection was conducted. A system dubbed the Wireless and Wearable Event detection and Adherence Monitoring System (WEAMS) was developed to explore the data that could be captured from the neck for the purpose of detecting and differentiating between swallow actions and other activities of daily living. In this chapter, a functional prototype of the WEAMS is presented, developed making use of off the shelf components. An experimental data collection was then performed to highlight the capability to acquire continuous multimodal data.

3.1 System Overview

The WEAMS comprises of multiple sensors in an integrated wireless wearable architecture for monitoring of health related signals. The system is meant to be worn on the neck in order to take advantage of many measurable physiological signals from respiratory, nervous, digestive, and circulatory as well as inertial readings of the body, continuously and noninvasively. The focus of this design is for collection tracheal signals wirelessly for evaluation of signal properties to be implemented in a real-time swallow detection system. To this end, a system was developed to use microphones and

a flex sensor for tracheal recording, capturing both audible data and the inertial movement from neck during various activities.

The prototype was based on the Sensortag[®] platform, which is an off the shelf sensor device with Bluetooth communications developed by Texas Instruments' (TI, Dallas Texas) [90]. This platform includes several sensors (gyroscope, accelerometer, humidity, temperature, and pressure) and BLE communications. The board contains the low power TI CC2541 microcontroller unit (MCU), which has on-chip BLE capability. This board was used to capture inertial data and provides an ADC through the CC2541. Additionally, a Bluetooth headset from Jabra was used for the transmission of wireless audio signal to the computer for data storage. Together, the Jabra Bluetooth headset board and the SensorTag were used for transmission of all wireless data. A block diagram of the WEAMS architecture overview is shown in Figure 6. Components of this prototype can be seen in Figure 5

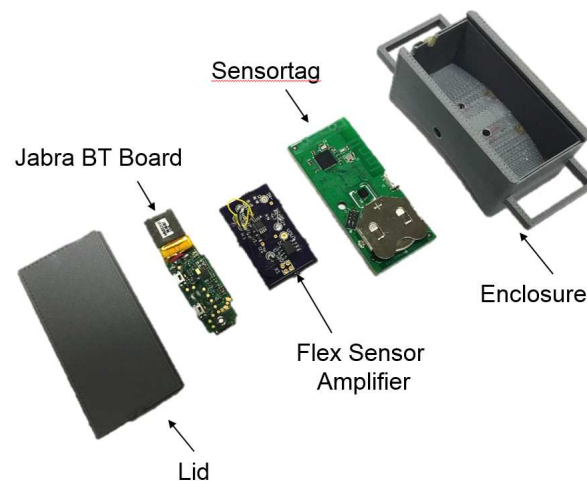


Figure 5. Hardware components of preliminary neck-worn prototype (WEAMS).

Tracheal activity recording is accomplished using a combination of two throat microphones and a piezoelectric flex sensor. The iAsus NT3 throat microphone was used to capture audible sound from throat [91]. This microphone is a contact type microphone aimed for recording only throat sounds and not ambient noise. In this prototype, one microphone was interfaced with the Bluetooth headset board to digitize and transfer the audio signal via Bluetooth. The Bluetooth headset provides some background noise filtering and filters out acoustic signals above the 4 kHz frequency range. The other microphone was connected with an audio output jack through a 3.5 mm mono audio connection and sampled by the PC sound card at 44.1 kHz with 16-bit resolution for a high quality comparison to wireless sampling.

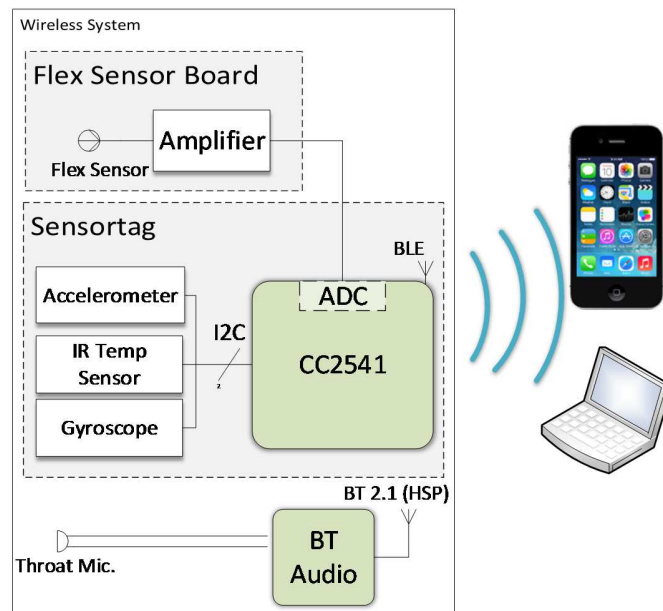


Figure 6. Block diagram of the WEAMS platform based on the TI Sensortag and Jabra Headset.

The WEAMS prototype used the DT-1 (Measurement Specialties, Hampton VA) laminated piezoelectric flex sensor , which converts the movement of the sensor into a voltage on two differential nodes [92]. For collection of the flex sensor data, an instrumentation amplifier (INA333) was used to condition the piezoelectric circuit so that it would be compatible with the dynamic range of the ADC input. The amplifier output was connected to the CC2541 by making a modification to the SensorTag board so that an open ADC input pin on the MCU could be accessed. The signal from this sensor was biased and amplified with a gain of 1.5 before it was sent as an input into the 12-bit ADC.

BLE was used for transmission of the data from the SensorTag to the Computer. The SensorTag firmware was then modified to allow transmission of the ADC output and the battery level. The SensorTag was programmed as a GATT server in which each sensor was assigned a universally unique identifier (UUID). Each UUID could then be requested by the base station and set to transfer data with regular notification intervals. Wireless tracheal audio signal was transferred over the Bluetooth headset profile (HSP) which provides an audio link with 16 bit resolution and 8 kHz sampling rate between the Jabra and the base station computer. This audio could then be accessed similar to a microphone from the PC. A software block diagram of the firmware and receiver software for data transmission is shown in Figure 7. Block diagram showing the software interface between the SensorTag firmware and the receiver. Represents a BLE GATT server.

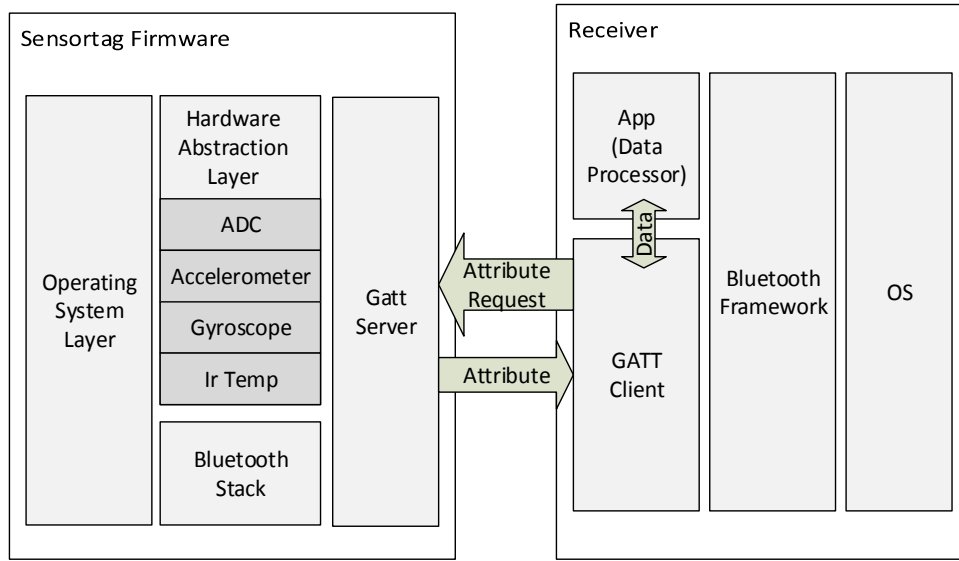


Figure 7. Block diagram showing the software interface between the SensorTag firmware and the receiver. Represents a BLE GATT server.

3.2 Data Collection Methods

For collection of tracheal and inertial data from the neck, the WEAMS system was tested on subjects through a study approved by the Georgia Institute of Technology Institutional Review Board (IRB). The two microphones are positioned on each side of the trachea, while the piezoelectric flex sensor is positioned slightly below the laryngeal prominence to detect throat movements associated with swallow actions. The synchronized wired and wireless recording of microphone data was for a quality comparison purpose. The components were placed in a 3D printed box and placed on the back of the neck. The whole system is secured using an elastic band from the throat microphones and is fasted with a magnetic connector. Figure 7a shows the WEAMS system placed on the neck. Data from sensors connected via SensorTag was sampled at

10Hz for each sensor. Bluetooth audio was sampled at the default Bluetooth standard connection of 8kHz with 16bit resolution.

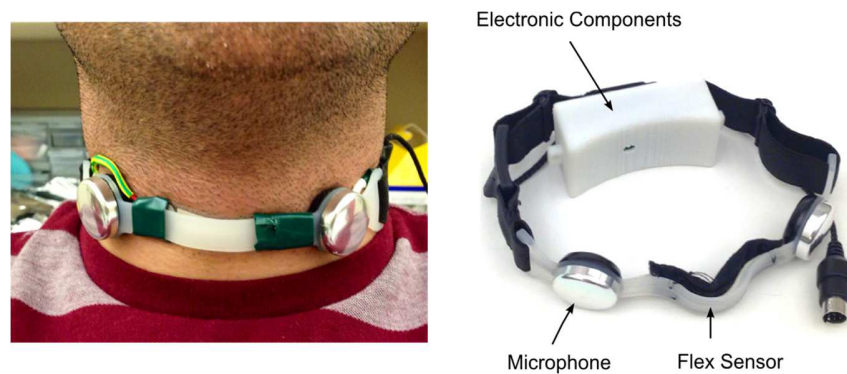


Figure 8. WEAMS neck-worn system. Device being worn on the neck with correct microphone placement on each side of the trachea (left). Overview of device components (right).

The study consisted of several phases over two days in which subjects were asked to perform a predefined list of tasks while wearing the WEAMS prototype in a laboratory setting. Items consumed in this experiment include water, orange juice, yogurt, gelatin capsules and crackers in order to cover a variety of bolus sizes and consistencies. Data was recorded using an interface developed with C# on a PC using the Windows 8 platform. The interface program used the BLE stack in Windows for collection of the sensor data and provided graphing and recording of each sensor. The subjects performed the action listed on the screen with the provided materials (water, OJ, crackers, pills) and could navigate between tasks using the “next” and “previous” buttons. This interface for the human subject trial is shown in Figure 9. One microphone connected to the onboard

soundcard (44.1kHz, 16bit) of the computer for quality comparison to the wireless audio. The gain of the Bluetooth audio and wired were matched so that signals gave similar amplitudes and such that tracheal sounds could be easily heard. Temperature data was also put on and removed the prototype. This task was included to determine the transient response of the IR temperature sensor and to assess its use in temperature detection of the subject. An exemplar set for the various activities and their resulting sensor signals can be seen in Figure 10.

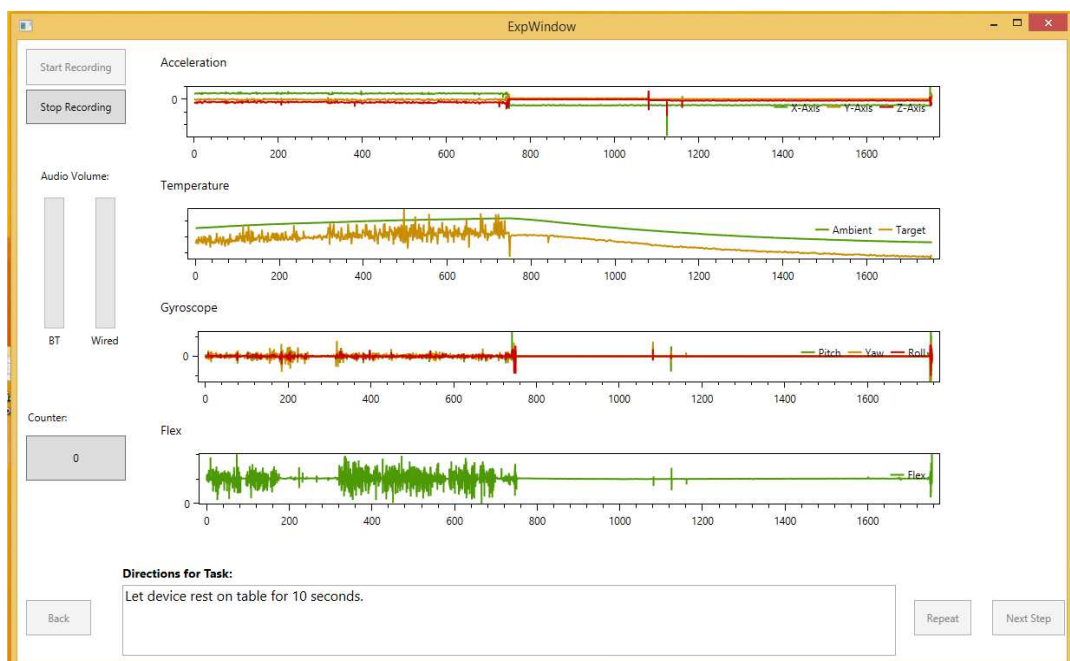


Figure 9. Experiment graphical user interface displaying task and recorded data.

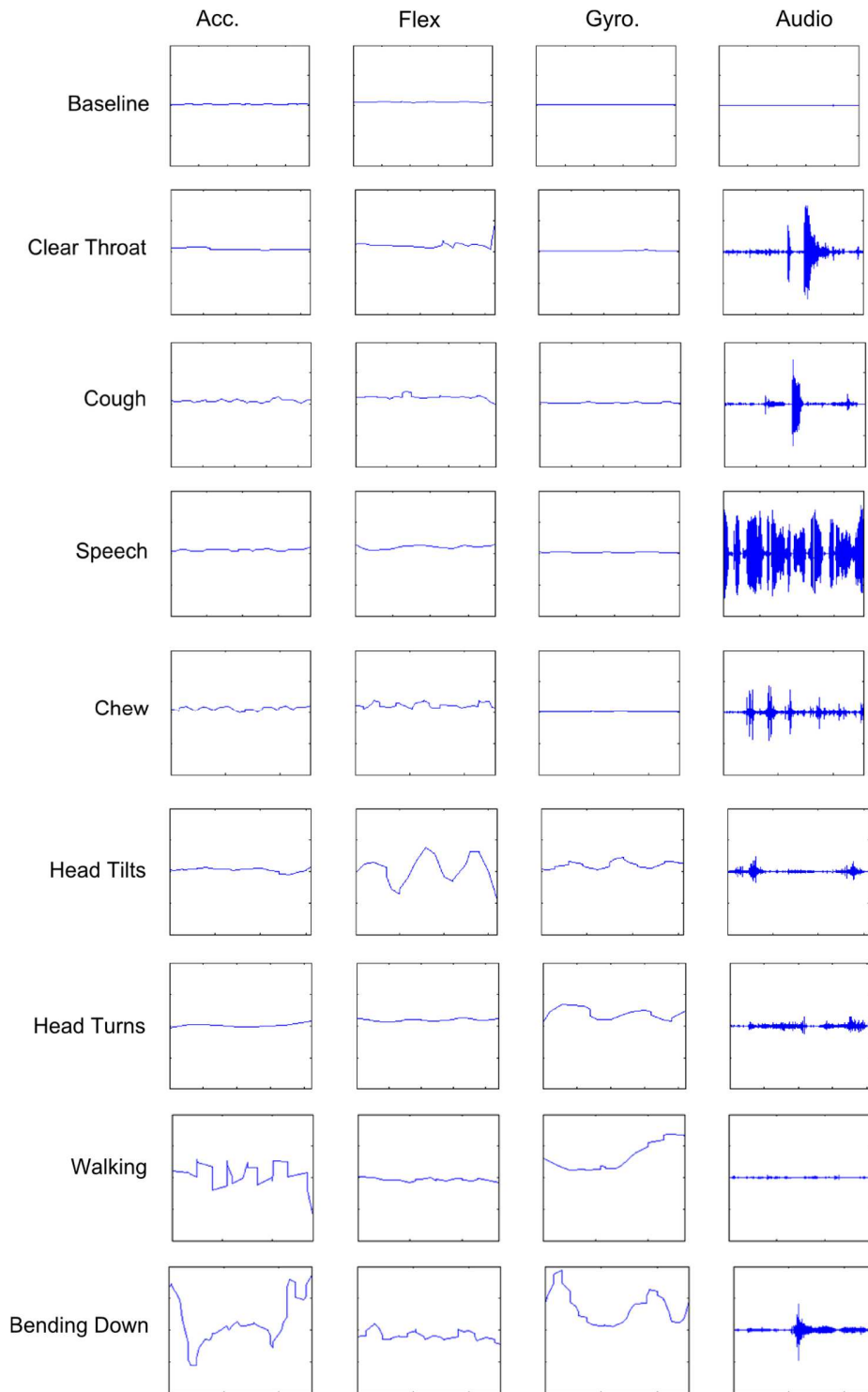


Figure 10. Recorded signals (accelerometer, flex sensor, gyroscope and Bluetooth audio) normalized between (1, -1) and sectioned to 5 second windows.

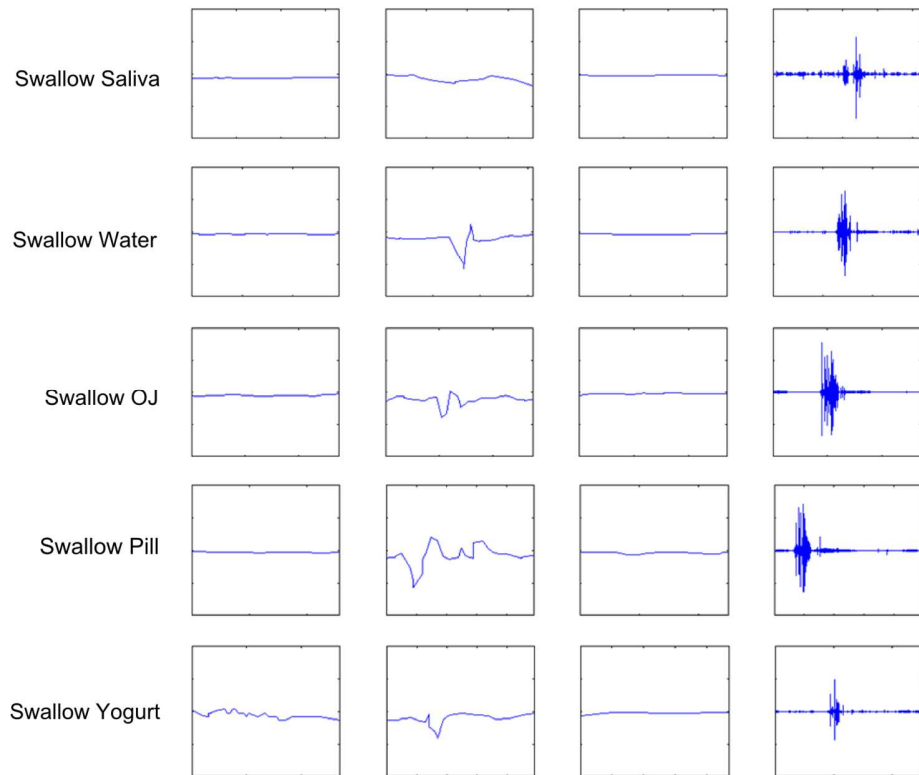


Figure 10 (continued)

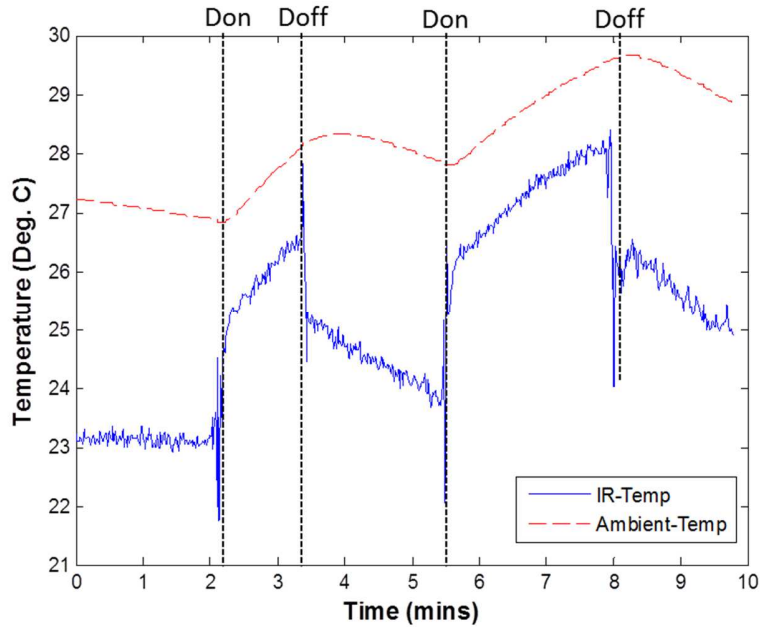


Figure 11. Temperature readings from the IR temp sensor including ambient temperature and object temperature.

3.3 Data Analysis & Results

Each activity was first validated and extracted from the data set using annotation labels created by listening to the audio signals with the exception of inertial signals, which were extracted using the known time of the experiment and expected sensor changes. These activities were then separated into five second windows, with the action centered within this window. Five second windows were chosen to match the time that the subjects were asked to rest between each action, so the data could be centered between rest periods. The magnitude of the inertial data was taken to extract the overall change in signal. For the accelerometer data, the gravitational acceleration was removed so that the signal would be centered at zero magnitude. After the data was extracted, the

signals were normalized to a range of -1 to 1 and the energy of each signal was calculated using the following formula:

$$Window\ Energy = \frac{1}{f_s} \sum S_n^2$$

where S_n is the sampled value from the sensors and f_s is the sampling frequency. The energy of each signal was then compared to a baseline, which was taken while the neck-wear was being worn, but no activities conducted. For comparison, a ratio was taken of the signals over the baseline value and converted to decibels for better presentation of large numbers. Table 1 shows the results of this analysis for one subject with the data represented as a multiple increase from the baseline signal.

From this analysis several observations can be made. The non-swallow tracheal activities (chewing, speech, cough, and clear throat) show similar patterns of having relatively low signals inertial signals and have smaller energies on average than that of the swallow signals. The most distinctive signal for the non-swallow group is the large energy shown in the audio sensing. This demonstrates that audio threshold may be the best method of discrimination when using the energy as a feature. Visually in Figure 10, it can be seen that the signal also varies in its composition indicating that the use of another feature may be able to distinguish between these signals.

The swallow actions show very similar energies in each signal category. The exceptions being empty mouth swallow and pill swallowing. The lower flex sensor energy of the empty mouth signal indicates that larger boluses have a larger signal energy using this sensing modality. The pill swallow showed higher energy in the flex sensor due to the greater difficulty subjects had in swallowing. This led to an unnatural swallow which can be visually seen with more activity in the plot of the flex sensor data for this

activity. Also shown visually is a similar waveform between each of the swallows. This indicates that the swallow follows a similar acoustic pattern and a frequency analysis may be able to identify swallows from other activities.

For the inertial activities (walk, head tilt, head turn, bend over), we see an obvious increase in inertial sensors indicating the inertial sensors are able to get signal from these actions from the neck position. Gross movements such as walking or bending over which require whole body motion have larger energies sensed by the accelerometer while the gyroscope shows to pick up small changes in orientation well. The head tilt data also shows a larger audio signal due to the rubbing of the microphone across the skin.

3.4 Discussion of Results

This preliminary study using the WEAMS prototype shows the ability to record data from human body with a wireless and wearable system. The data shows that signals show up differently for each type of sensing modality. For example, while head turns had similar energy levels for the flex sensor and BT audio, gyroscope signal energy was much higher for the head turn. This method provides some intuition about which signals should be combined for more accurate detection and discrimination of different activities. While energy of the signal was used in this preliminary, other features be more valuable than energy alone. Figure 10 shows very similar waveforms for swallow activities in the flex sensor and in audio. Other features such as rate of change or frequency analysis could be used to develop an efficient and accurate multi-modal sensing algorithm.

Table 1. Signal Energy for Each Signal and Sensing Modality as Percent of Baseline.

Activity	Accel. (dB)	Flex (dB)	Gyro.(dB)	Audio (dB)
Baseline (No Activity)	0.00	0.00	0.00	0.00
Chewing	9.21	7.95	5.73	20.96
Speech	5.76	9.10	1.56	34.15
Cough	11.27	8.69	8.19	25.52
Clear Throat	9.57	3.17	0.23	24.09
Empty Mouth Swallow	7.15	6.91	0.19	13.24
Swallow Water	2.49	10.58	3.74	16.42
Swallow Yogurt	9.51	10.67	1.68	9.82
Swallow Orange Juice	6.21	10.41	2.74	18.60
Swallow Pill	2.00	12.73	5.75	20.36
Turn Head	9.91	9.15	20.58	7.26
Tilt Head	8.70	4.59	15.89	16.62
Walk	20.86	1.33	26.05	3.96
Bending Over	23.75	6.63	26.38	13.51

CHAPTER 4

HARDWARE DESIGN OF A NECK-WORN HEALTH MONITORING SYSTEM

A custom hardware solution of a neck-worn monitoring system was designed to provide monitoring of several vital signs from a neck placement. This design was developed in order to provide expandability and more sensing components for health specific applications than off the shelf designs offered. The requirements of this design included the sensors: PPG analog front end, 9 axis inertial sensing, temperature sensor and the inclusion of an instrumentation amplifier front end for capture of other bio-signals. The implementation criteria of the design should implement the BLE protocol for transmission of data and should be relatively small ($< 2'' \times 3''$). The design should also include battery charging for ease of use. A block diagram displaying the components of the design can be seen in Figure 12.

4.1 MCU and Bluetooth design

The Texas Instruments (TI) (Dallas, TX) CC2541 integrated MCU and Bluetooth Low Energy IC. The CC2541 makes use of an 8051 architecture as the microcontroller core. This MCU was chosen for its on chip BLE capability and low power requirements. The MCU also provided both the I2C and SPI serial communications protocols allowing a wide variety of sensors to be connected to it. Additionally, this MCU was chosen due to familiarity with the firmware development environment. The chip also comes with a Bluetooth stack provided by TI allowing for rapid development.

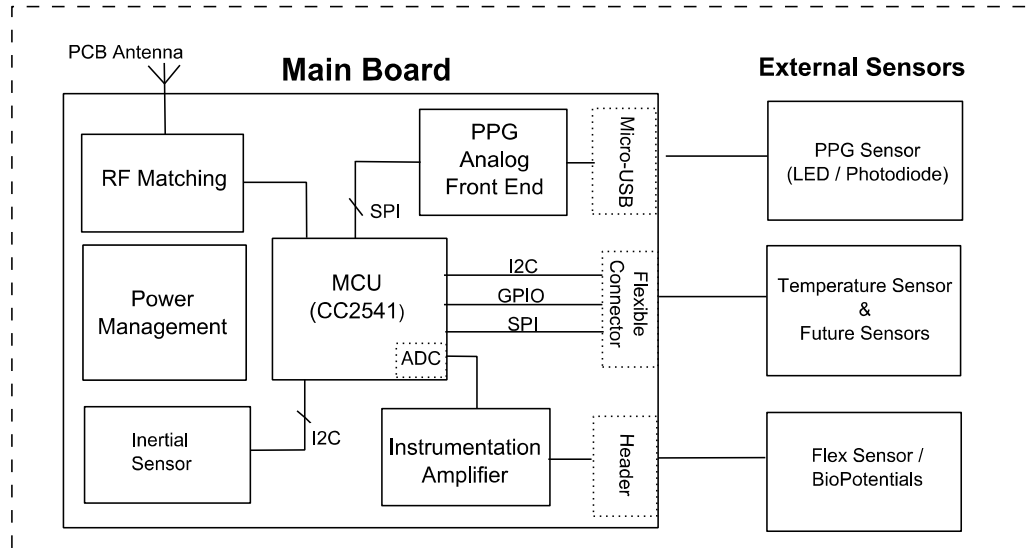


Figure 12. WEAMS custom hardware block diagram.

The schematic for the MCU portion of the design is shown in Figure 13. The MCU in this design is configured for six serial peripheral interface (SPI) connections two of which are on board and 4 are available through connectors for future expansion. In addition to the SPI connections, I2C connections are also available through an expansion connector for connection of ICs supporting this protocol. There is also provision for a maximum of five ADC inputs one of which is used for conversion of signal from the instrumentation amplifier front end. The designs expandable connectors provide power and both I2C and SPI protocols to additional off board sensors.

For Bluetooth capability, the RF component of the board was optimized for the 2.45GHz frequency. The MCU provides a balanced RF output for the antenna, so a balun was used to convert this output to an unbalanced output for the antenna. A balun was used instead of a passive component design in order to save space and effort in soldering.

The Johanson (Camarillo, California) 2450BM15A002 balun was chosen due to its specific impedance matching design for the CC254X series Bluetooth chipsets. This balun has an unbalanced output impedance of 50Ω for transmission line matching. A PCB antenna design provided by TI reference designs was used for BLE transmission and reception. The PCB antenna was chosen due to better performance and easier implementation than a chip antenna, requiring only an additional copper trace during manufacturing.

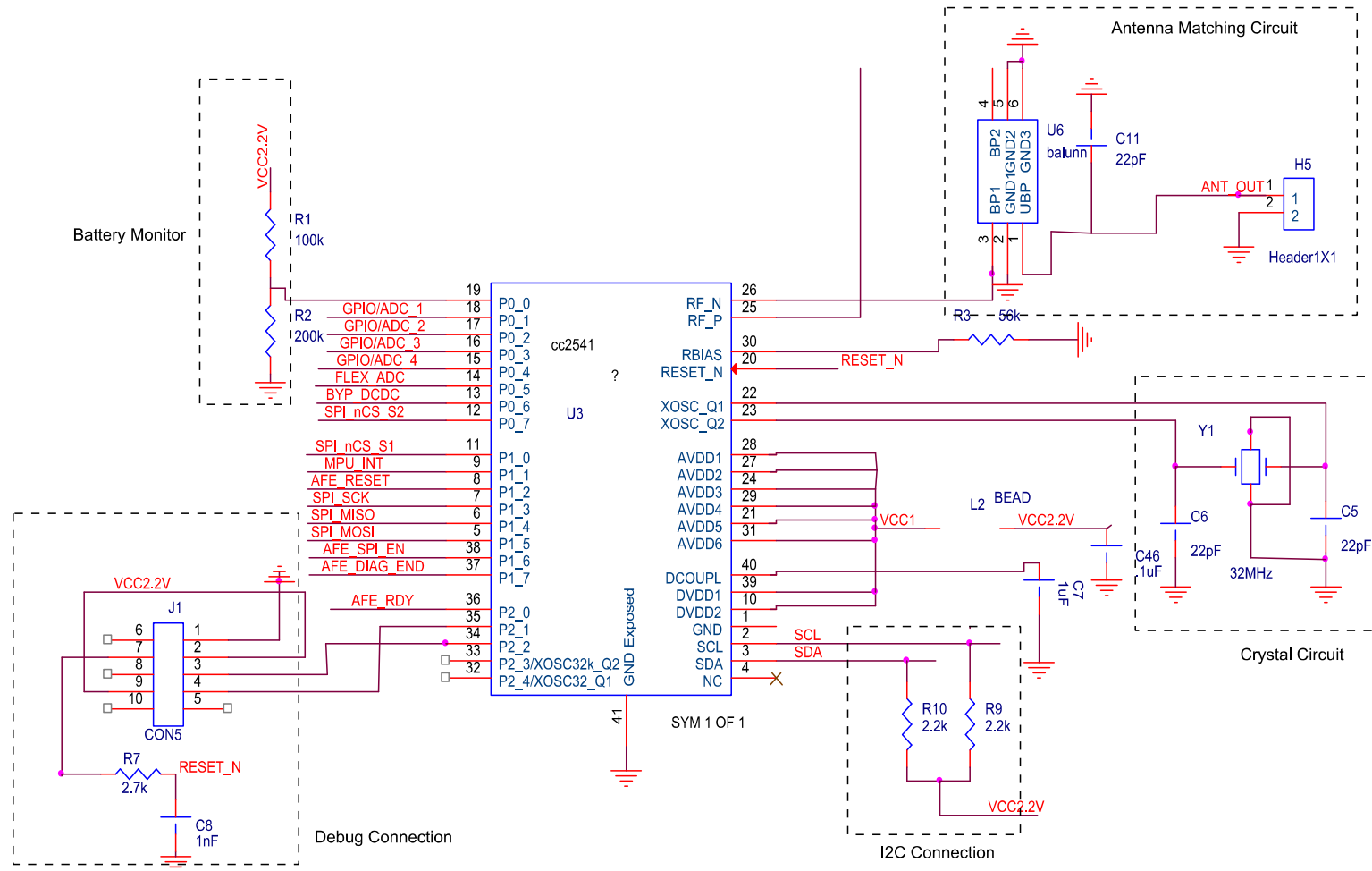


Figure 13. CC2541 MCU schematic including debug connector, communications ports, and antenna matching circuit.

The antenna topology implemented is inverted F with multiple folds with an approximated efficiency of 68% [93]. The layout of the antenna has an area of 15.2 x 5.7mm. Figure 14 shows the PCB footprint layout of the antenna.

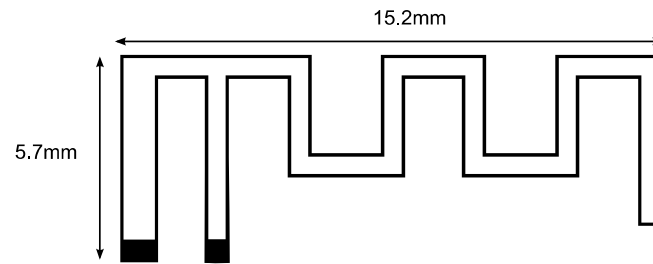


Figure 14. PCB antenna design layout.

4.2 Sensors

4.2.1 Inertial Sensor

For inertial sensing, the MPU-9250 (Invesense, San Jose) was used [94]. This sensor is a 9-axis digital compass containing a 3-axis accelerometer, 3 axis gyroscope and a 3-axis magnetometer for accurate detection of orientation and direction in 3D space. The MPU-9250 was chosen for its small size (3mm x 3mm) high precision, and low power consumption (9.3 μ A). The accelerometer scales from $\pm 2g$ to $\pm 16g$ while the gyroscope supports from ± 250 to ± 2000 ($^{\circ}/\text{sec}$) in all orientations with 16bit resolution. The magnetometer has a full scale range of $\pm 4800\mu\text{T}$ and 14 bit resolution. These limits provide sufficient range for detection of inertial activity including large signals such as falls. I2C and SPI are both supported by this chip design but I2C was chosen as the

interface in order to limit the number of pin connections on the PCB. Figure 15 shows the schematic of the MPU-9250 interface.

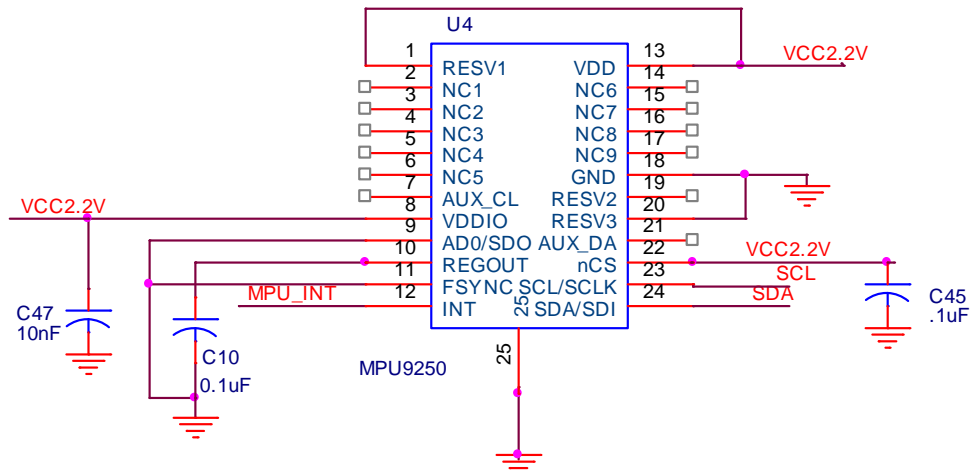


Figure 15. Application schematic of MPU-9250 digital compass.

4.2.2 Temperature Sensor

The temperature sensor is located on a separate board in order to position it closer to the skin of the wearer. This board is connected through a flexible ribbon cable for the purpose of mounting in an advantageous location for collecting the temperature. A Texas Instruments supplied IR temperature sensor (TMP006) was used so that direct thermal contact would not be necessary between the wearer and sensor. The TMP006 also includes an ambient temperature sensor for sensing of the environment. The IR sensor provides a range of 0° to 60° Celsius range, which is adequate for the detection of skin temperature and is accurate within 3 degrees. The ambient temperature sensor in the TMP006 is accurate to ± 1 degree Celsius with the same range as the IR temperature

sensor. The TMP006 has an I2C interface for connection to the MCU. The schematic for the temperature sensor board is shown in Figure 16.

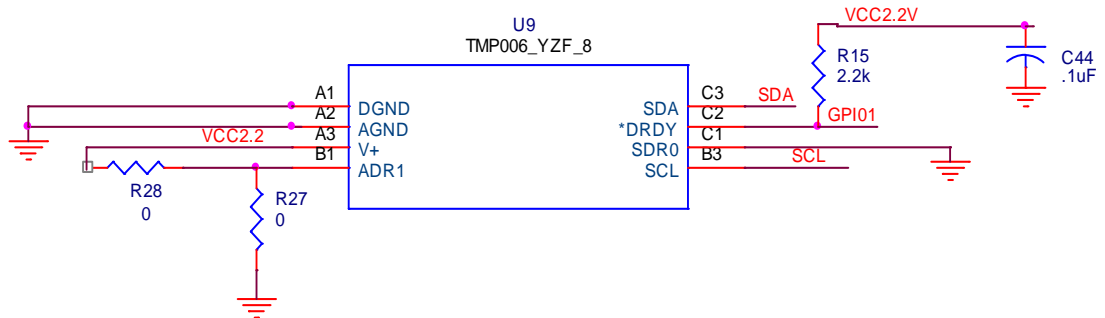


Figure 16. Temperature sensor circuit with connections to MCU shown. Here it also includes a 0-Ohm resistor for address adjustment.

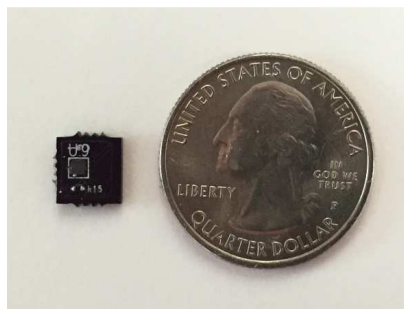


Figure 17. Populated temperature sensor board.

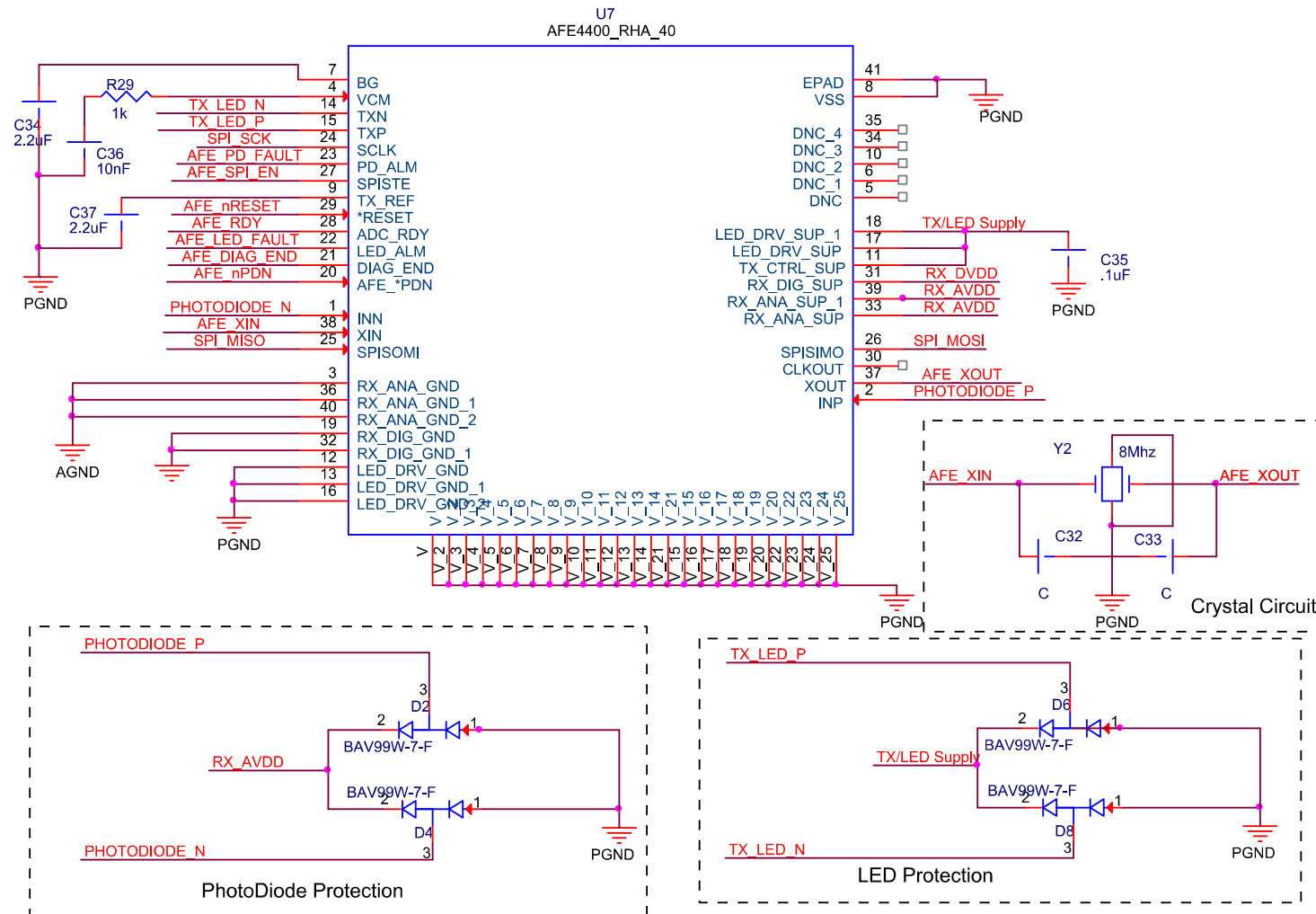


Figure 18. AFE4400 schematic displaying analog and digital connections, protection circuit, and crystal.

4.2.3 Pulse Plethysmography Sensor

4.2.3.1 AFE

The PPG sensor consists of two parts: the analog front end (AFE) located on the main board and the PPG LED and photodiode located on a separate board. The two parts are connected via a shielded USB micro connector, chosen for its small size and adequate number of conductors. The AFE used in this design is the AFE4400 (Texas Instruments, Dallas, Texas), an integrated chip solution for PPG applications [95]. The design also supports the AFE4490 which has higher resolution detection, however this was not considered necessary in the initial design as the power requirements and cost higher. [96]. This AFE provides several capabilities including, LED drivers, filtering and amplification of the received signal from the photodiode, a bias removal circuit for lowering noise due to ambient light, and an ADC. The AFE offers an 8 bit programmable LED driving current with a range up to 50mA. Both heart rate PPG and full support of pulse oximetry sensing is supported with this AFE and is dependent on configuration of the optical transceiver.

The AFE is interfaced to the MCU using SPI as well as general purpose IO pins, which provide hardware interrupts to the MCU in the occurrence of an LED fault and ADC conversion completion so that errors in detection can be handled by the MCU firmware. The schematic for the AFE4400 is shown in Figure 18.

4.2.3.2 PPG Sensor Board

For the PPG detection board board, a combined LED array and photodetector in single package provided by APM Korea (Daejeon, Korea) [97]. This detector was

chosen for its small size of 4.8mm x 2.6mm x 1.0mm and power requirements being within the range of the AFE. The DCM05 contains an IR LED with peak emission of 905nm and a red LED with peak emission of 660nm as well as a photodetector with a range covering these two wavelengths. The design is setup in H-Bridge configuration for driving the LEDs. Figure 19 shows the dimensions of DCM05 device as provided by APM Korea and Figure 17 shows the board compared to a US quarter.

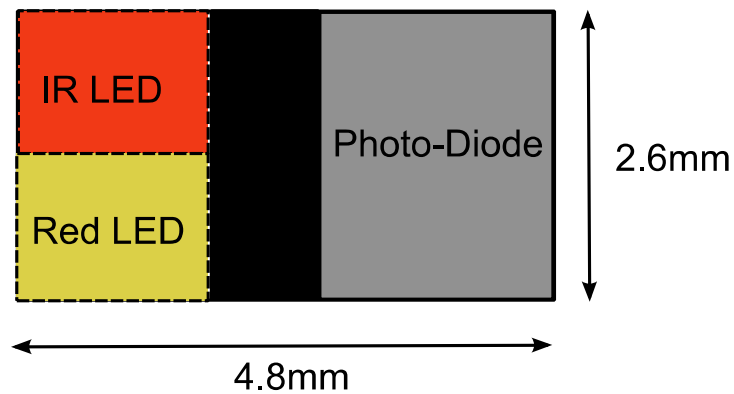


Figure 19. PPG sensor board from APM Korea with two LEDs and a photodiode integrated into one package.

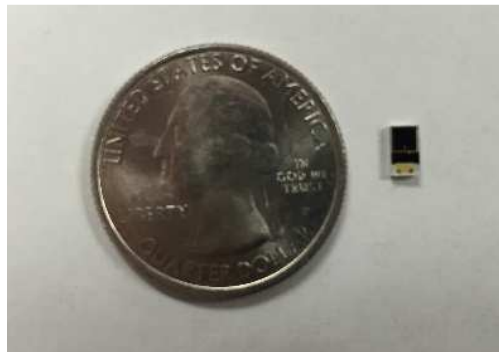


Figure 20. PPG sensor board with coin for size comparison.

4.2.4 Instrumentation Amplifier Analog Front End

An instrumentation amplifier was included in the design for amplification of biosignals or signals provided by piezoelectric force sensors. The instrumentation amplifier is used to condition the signals to fill the full dynamic range of the ADC input of the MCU. Though the amplifier may accept a broad variety of signals, in this case it was designed for two different applications which are chosen during board population. These applications are a piezoelectric flex sensor or the future use of EMG. This amplifier circuit is designed to amplify and center signals in the range of 0 to VCC as accepted by the ADC. The INA33 IC (Texas Instruments, Dallas, Texas) was chosen for the instrumentation amplifier [98]. This amplifier has a low supply voltage requirement of 1.8V allowing it to be powered with the MCU power supply circuit, and operates on a single power rail. An instrumentation amplifier was chosen over an differential amplifier design due to a higher input impedance to match the high impedance output of the desired signals and also due to a very low noise and high common mode rejection (CMRR), which in this case are 50nV per $\sqrt{\text{Hz}}$ and (100db) respectively [98]. The amplifier has a hardware adjustable multiplier provided by a single resistor with an allowable range of 1x to 1000x amplification. Additionally, a reference voltage of VCC/2 is provided so that the output signal of the instrumentation amplifier is biased to the center of the ADC voltage range. This allows negative and positive signals to be detected after the bias is removed in post processing.

4.3 Power Management

This design employs multiple power management units in order to supply provide efficient power supply to the MCU and analog front end. The power supply is designed for a lithium Ion battery with a nominal voltage of 3.6V. Both power supply pathways use this battery source as the power input. A battery charging circuit is also included in the design and operates using the USB standard (5V, 500mA) power input. This design allows the device to be charged with off shelf USB micro chargers which are readily available.

4.3.1 Charging

Figure 21 shows the charging circuit implemented in this design. This charging circuit uses a Linear Technologies (Milpitas, California) LTC 4054[99]. This is a linear charger IC providing charge to Lithium Ion chemistry batteries with 3.6-3.7V nominal voltage. This chip was chosen for its specific compatibility using USB power supplies. The LTC4054 is capable of providing 800mA of current to and allows variations in voltage supply from 4.25V to 6.5V. Current programming is controlled by the “PROG” pin input using a resistor to set the value to a level that is suitable for the battery. Three modes of operation are supported in the normal charge cycle. At very low charge the battery is first supplied $1/10^{\text{th}}$ the programmed current in a trickle charge mode. After the battery reaches 2.9V the constant current mode is used to bring the battery to 4.2V using the programmed current. At 4.2V the charger enters constant voltage mode and ends the charging cycle when the current is lowered to $1/10^{\text{th}}$ the programmed current.

4.3.2 MCU Supply

The CC2541 contains an on chip low dropout linear DC voltage regulator (LDO) for producing a voltage of 1.8V for chip operation. However, the efficiency of an LDO

depends on the battery voltage and is less efficient for higher battery voltages when the battery capacity is full. The LDO performs reasonably well at lower current consumptions such as when the RF radio is not active. A DCDC voltage regulator provides better efficiency during the higher current draws during RF communication. The drawback of using a DCDC converter is that the power conversion may be less efficient at very low current consumptions. Also a DCDC converter must be carefully picked in RF applications due to a resonance frequency caused by the DCDC switching circuit. In this design a TPS 62730 (Texas Instruments, Dallas, Texas) DC to DC buck converter was included to supply the MCU during RF transmission. This DCDC implementation is shown in Figure 22. This DCDC converter has low noise in the Bluetooth frequency range and has been shown to not interfere with the operation of the CC254x series [100]. The inclusion of this DCDC converter allows for over 30% reduction in current consumption from the battery during RX and TX when the battery is in a fully charged state of 3.6V [100]. For maximal efficiency the DCDC converter is shut down during periods of RF silence. This shutdown of the DCDC converter is software controlled by the CC2541 so that it may be coordinated with the Bluetooth data transfer.

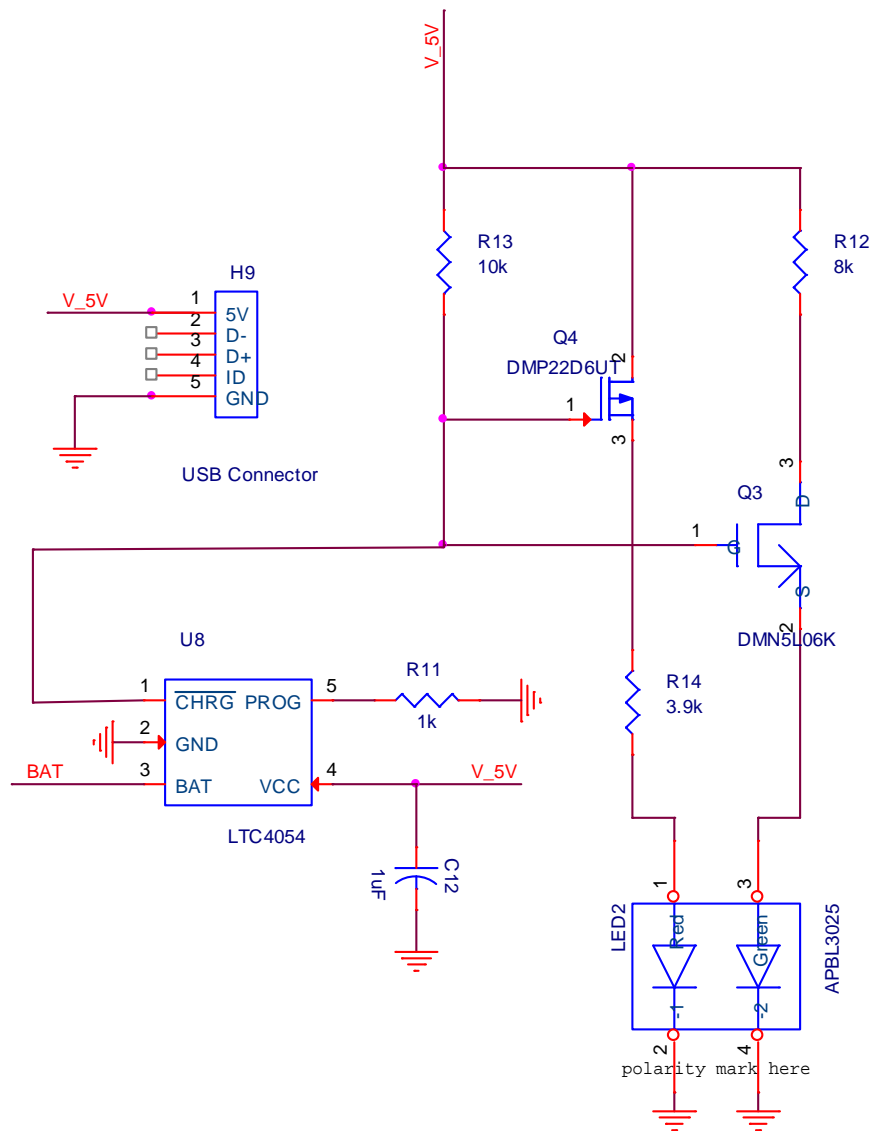


Figure 21. Charging circuit with LEDs for indicating charging status.

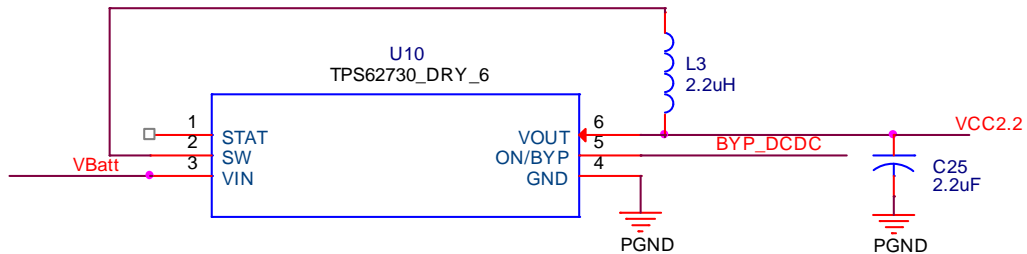


Figure 22. DCDC converter circuit for supplying of the MCU and other digital sensors.

4.3.3 AFE Supply

Due to the requirements of the PPG AFE, an additional supply must be used. The AFE 4400 requires a power supply range of 2V to 3.6V for the receiver supply voltage and a voltage of 3.0V to 5.25V for the LED transmitter supply. The LED transmitter supply used depends on the LED application, and a higher voltage may be needed for accurate detection of PPG signal. A DCDC boost regulator was coupled to two secondary LDO regulators each with adjustable gain in order to meet different supplies of the RX and TX channels. The design also includes the ability to bypass one of the LDOs is the voltage required for the LED transmit signal lies within the voltage range for the receive supply.

The first stage of the power supply for the AFE is a DCDC boost converter. The boost converter was used so that the battery could discharge below the 3.0V level and provide the voltage required to be in the operation supply limits. The TPS61093 (Texas Instruments, Dallas, Texas) boost voltage converter was chosen for its ability to supply

5V with a low minimum input voltage of 1.6V [101]. This DCDC converter has up to 88% efficiency and an adequate 1.1A peak current limit. The DCDC converter has an enable pin that is tied to the MCU to allow shutdown of the AFE power supply for implementing a low power mode.

After the DCDC converter there are two parallel LDO stages. LDOs were chosen for this stage because the voltage provided by the DCDC is close to the output voltage of the LDO limiting inefficiency in regulation. The operating mode of LDOs also provide a AC frequency filtering effect which limits the noise from the DCDC switching frequency and provides a cleaner signal for the DC analog lines being supplied to the PPG AFE. This trait was desired so that the switching frequency of the DCDC would not effect measurements of the AFE due to noise on the supply inputs. The TPS7A4901 (Texas Instruments, Dallas Texas) LDO was chosen as the supply regulator due to their low noise properties [102]. This LDO has a resistor adjustable output range from 1.19V to 33V from a 3V to 36V input and a 150mA current limit. The TPS7A4901 has a power supply rejection ratio (PSRR) of 72dB and is designed for the application for sensitive analog circuitry. These LDOs may also be bypassed based on configuration needs. The entire AFE supply schematic is provided in Figure 23.

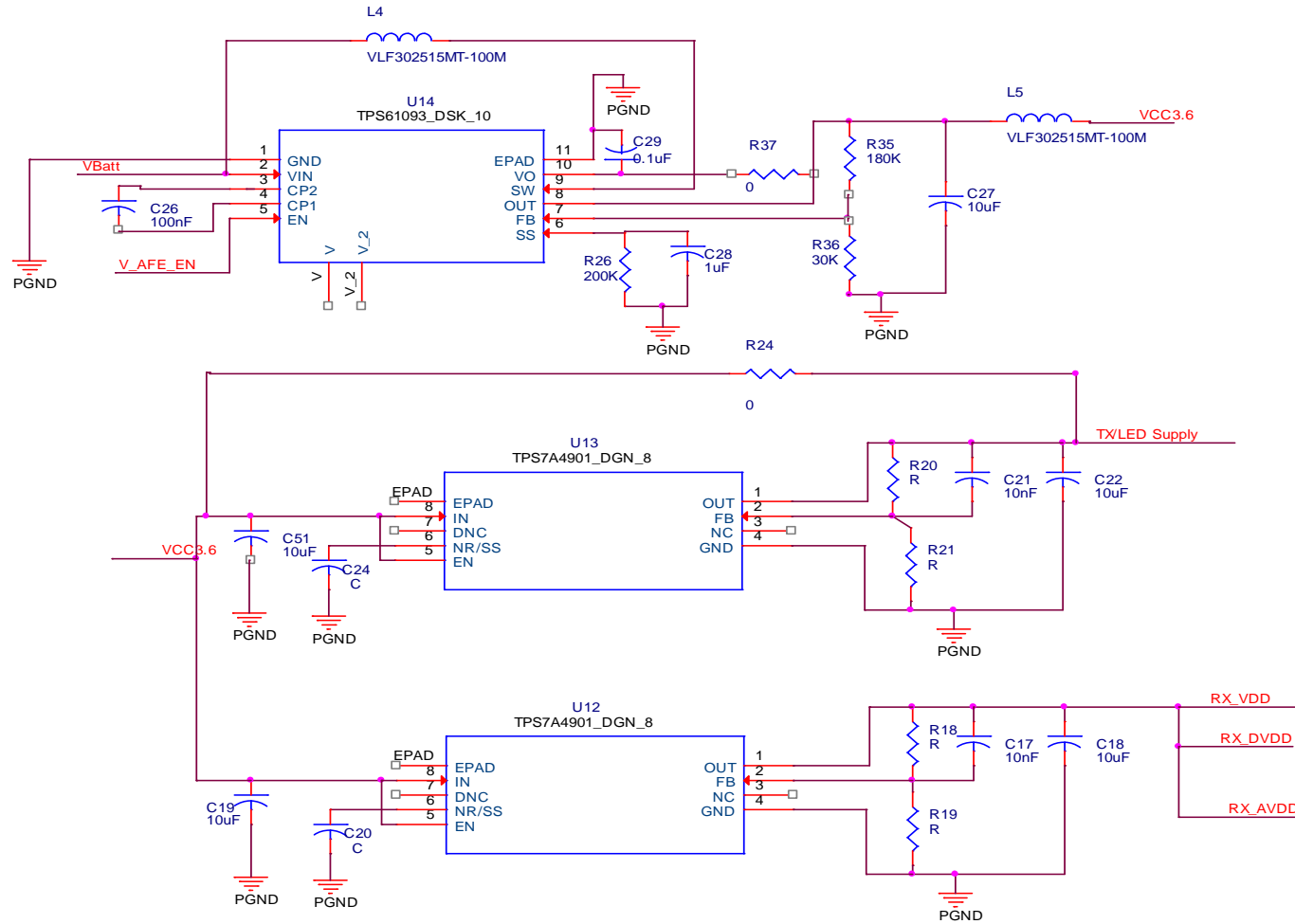


Figure 23. AFE power supply blocks. TPS61093 provides boosted voltage and TPS7A4901 each provide regulated voltages and blocking of DCDC noise. $V_{CC3.6}$, TX/LED, RX VDD, RX DVDD (digital), and RX AVDD (analog) are each provided in this circuit.

4.4 Results

The final board produced resulted in a size of 39mm x 21.8mm x 4.5mm. This met the initial size requirement goal and is a reduction in size from the initial prototype. Figure 24 displays the new design in comparison to the Sensortag based prototype hardware. The final specifications of the design may be seen in Table 2.

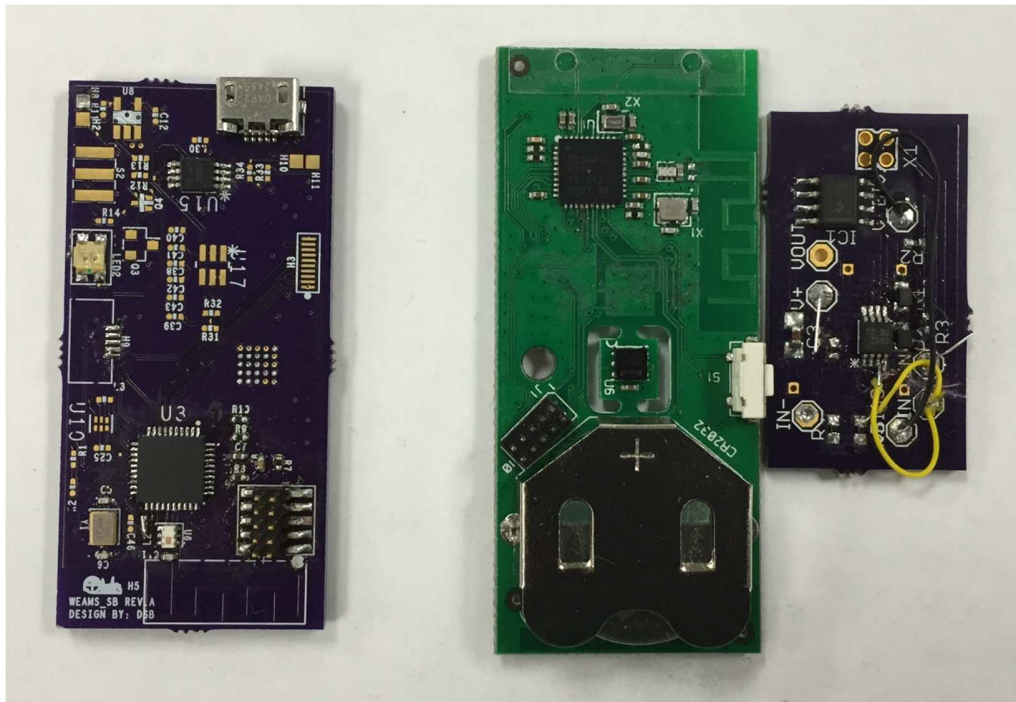


Figure 24. Comparison of custom hardware (left) and first prototype board (right)

Table 2. WEAMS Hardware Specifications

Board Size	1" x 2" (24.4mm x 50.8mm)
Battery	3.7V Lithium Polymer Charging Capability
Power Consumption	22mA, (Duty Cycle Usage)
Magnetometer	4800uT range, 14-bit resolution
Accelerometer	16g range, 16-bit resolution
Gyroscope	2000°/s range, 16-bit resolution
Instrumentation Amplifier	1-1000 adjustable range, 12-bit resolution
PPG Sensor	22-bit single channel reading, Pulse Oximeter Compatible

4.4.1 Battery Life Estimation

Battery life was estimated using the known current consumptions of the sensors and overhead of circuits in the system. A target battery of 240mAh was chosen for the estimations for having a similar dimensions as the board of (39mm x 21.8mm x 4.5mm). This allows the board and battery to be fitted to a compact enclosure. Three estimations were performed: a max load estimation with each sensor in the on state continuously and the BLE link active at all times, a medium load state with all sensors running at 50% duty cycle, and a sleep state with all sensors in sleep mode and the BLE link inactive. Table 3 shows the time of battery life estimation for each power mode.

Table 3. Battery Life Estimates

	Max	50% Duty Cycle	Sleep
Current (mA)	43.9	1.6	22.4
Time (hours)	5.47	148.05	10.718

CHAPTER 5

DISCUSSION AND FUTURE WORK

5.1 Discussion

The ideas presented in this work propose a novel multimodal neck-worn device for use in health monitoring. The need for a pervasive, wearable monitor is growing due to an increasing burden on the healthcare system which may be alleviated in part by wearable devices providing medical information. The prototype presented in this paper may be used to meet the growing needs for people engaging in independent living, while suffering from chronic conditions. There is a special for people with chronic conditions to have medical information logged for use by a physician or caretaker in preventative medicine. An implementation of a neck-worn device in this population could provide more independence and freedom through medical informatics. A health monitoring device that can detect swallows of food and medication may be beneficial to these populations including elderly and adolescent. This is especially applicable to those with severe chronic conditions whose chance of survival may depend upon a rigid medication schedule.

The work here has many applications in the healthcare space. A neck-worn system with its unique position for sensing of vital signs, may prove to be useful in specific medical research topics such as dysphasia or swallowing rehabilitation. A device that can be used for pill swallow detection is also useful for pharmaceutical studies, in which the manufacturer must be assured that the volunteer population is taking the medication in order to assess the effectiveness and possible side effects of proper

dosage. A device that could guarantee compliance to a medication regimen in this situation would allow the studies to be more robust and allow a wider pool of subjects if they can be monitored remotely.

This device builds upon the work presented by others providing an expandable platform in that allows multiple uses from one hardware design unlike other swallow detection designs that are such as one being developed at University of Alabama that relies on accelerometer and flex sensor from the jaw only [11]. The inclusion of multiple sensors also allows greater distinction of activities that might provide similar signals in a single domain of data collection. This provides a richer experience in comparison to other devices that use one modality for swallow detection such as the flex sensor based system developed at UCLA [82].

The use cases presented are only a few examples of how the work presented here could prove to be beneficial in the healthcare and medical research fields. There is ongoing work on this project with the goal of implementing these ideas and further building upon the designs presented with work being done in interface design, algorithm implementation and further design of neck-worn hardware.

5.1.1 Human Factors

A wearable system must provide enough benefit to outweigh the drawbacks of donning the system. Potential drawbacks are inconvenience, burden, and especially in the case of wearable devices in visible locations, industrial design. If the device has large deficiencies in these areas or fails to provide enough perceived benefit then it will not be worn, and the potential value of the health data is lost. In the introduction chapter, the need of wearable devices for applications in pHealth was shown and potential value of

these devices was shown subsequently. Since potential benefit is known, the industrial design of the system must not be focused on. To this end, preliminary work has been done in making the device presentable to elderly populations.

Two studies involving elderly populations were performed to assess the acceptance of a neckwear device. The first study involved a questionnaire presented to elderly peoples ranging 66-96 years of age after showing renderings of possible neck worn devices. The results showed that if instructed by a physician the majority (70%) would wear a neck-worn[7]. A later study performed in association with HomeLab [103], a Georgia Tech initiative for studying new devices in an elderly population, used fabricated mockups to assess the wearing of the device [104]. In this study, fabricated neck-worn mockups were presented alongside several commercial products (Jawbone Up, Mio Alpha, BodyMedia FIT). Impressions of the neck-worn device were recorded taken and recorded from the participants. The results showed that the subject population found a magnetic latching neck-worn system to be the easiest to use over snap and sliding arm types. It was also shown that the neck-worn devices were most intuitive in putting with the exception of the Jawbone UP [104]. Again in this population, it was shown that the subjects would wear the device if it was recommended by a physician. Interestingly, every subject indicated that they already wear some kind of device daily in the form of hearing aids and similar devices.

Though the results of these studies are promising for potential usage of neckwear devices, there is still a disparity between the mockups and actual device in terms of size and look. More work must be performed in making a form factor smaller and more manageable for ease of use and compliance by a wearer.

5.1.2 User Experience

Equally important to the appearance of the device is the usability. Wearable Health devices present a large opportunity for personalized healthcare and the ability for the wearer to gain insight about their own habits. The presentation of the health information must be easily accessible, and informative to the user. Smartphones provide a good platform for interfacing with wearable systems due to the inclusion of larger screens, expandable application platforms (iOS, Android) and inclusion of Bluetooth for wireless data transmission. Already Apple and Google have started implementing health applications in their platforms with the introduction of Google Fit and HealthKit™ [105], [106]. Future work should explore using these platforms for the presentation of medical data collected from the neck-worn system.

5.1.3 Privacy

There are many implications of a pervasive worn device for the wearer's privacy. A neck-worn monitor that records acoustics has the potential to capture sensitive conversations. This privacy issue can be mitigated by using algorithms which can distinguish speech from other activities so that this data will not be captured. Additionally, the collection and storage of medical data must also be considered. Providing real-time processing of the data and only saving statistics on the data can mitigate the issue of security of the storage of data. Privacy considerations for storage of medical data after processing must be handled using current protective technology, which generally require encryption of the data and some unique identifier for data access. This portion may be handled similarly to current medication information storage techniques in place today used for electronic health records.

5.1.4 Considerations for Continuous Signal Processing in a Wearable System

Because wearable devices allow little space for battery storage to keep their small form factors, special considerations must be made for a real-time processing system that can analyze and detect signals as they occur. This is especially true for audio processing because of the higher frequency sampling required to capture the signal and the increased processing requirements to perform analysis of the signal as compared to other relatively simple signals such as inertial signals. In order to meet these challenges, low-power, embedded digital signal processors (DSP) have recently been developed for the use of audio processing. Dedicated signal processing units are ideal because they are optimized for the function of audio processing and do not require the large overhead of a generalized processing core. These processors can also work locally analyze the signal without the requirement of server processing which is currently commonplace in voice processing systems such as Siri™ in Apple's smartphones and other similar systems. A major benefit of this is low latency between reception of the audio and trigger analysis for use in time-sensitive applications. The device is also able to act in a stand-alone fashion if a receiver is unavailable. One example is the Cadence Tensilica Hifi Audio DSP, which is a digital processing unit that consists of a lower power DSP for always-on listening as well as the capability for data fusion of other sensors. This provides voice trigger functionality that can wake up an external MCU after the utterance that matches the desired trigger audio [107]. Similar technology has been implemented in the Motorola MotoX™ smartphone. In the MotoX system, a low power DSP (C55) from TI was programmed to process audio data for detection of a voice trigger while reducing power consumption by 5 times of standard phone voice recognition [108].

In addition to the processing, analog filtering and amplification of the signal may also be optimized in order to provide a lower power consumption which is suitable for an always-on wearable device. To this end pre-amplification stages may also be optimized so that they can dynamically adjust to the bandwidth of the signal. Du et al have demonstrated such a system with adaptable bandwidth which can provide up to 84% of power savings without significant signal degradation of audio signals [109].

The technology developed for voice operation may be leveraged for development of devices meant to detect other signals involved with health monitoring. In this case, a specialized signal processing unit implemented either on a DSP or FPGA system may provide a real-time trigger for tracheal activities so that a person may be monitored using a standalone system.

5.2 Future Work

This work presents a preliminary assessment of a neck-worn system for tracheal activities and a design for a multimodal neck-worn health monitoring system. The results show promise for the neck-worn system and represent a novel location for collection of many types of vital signs from multiple physiological systems of the human body. Additional work is required to demonstrate a fully implemented neck-worn system. The medical field is rapidly changing, but thus far there are few systems that have been implemented specifically for personalized healthcare of sick populations with wearable devices. However, there are several considerations that must be explored and future work to be performed in this field before a neck-worn system may be presented to a clinical or consumer population.

5.2.1 Expandable Design

The design presented in Chapter 4 presents a platform that includes PPG, temperature sensing, Inertial sensing, and an instrumentation amplifier, but also the ability to be expanded with ports for additional sensors. This design intentionally presents itself as a platform, which may be expanded to accommodate other sensing modalities. Ongoing work will explore the sensors that may be placed at the ear, or elsewhere in vicinity of the neck for additional sensing capability. The expandable design greatly increases the value of the platform and will allow external sensing modalities to be quickly explored in a modular fashion. The ear provides an especially attractive expansion due to the additional vitals that can be provided accurately at this location. PPG and ear microphones are currently being designed for use in collection of blood oxygen and the onset of swallow respectively. These additional sensors may prove helpful in expansion of the neck-worn system.

5.2.2 RFID Pill Detection

One of the main project goals of the WEAMS system to be implemented is medication compliance through automatic pill detection. This will be accomplished through an RFID reader on the neck, which is activated during swallow activities. The detection of swallowing of items has been explored, but the detection of RFID pills entering the body has not been adequately studied. Work in preparation for this capability has been done, including the design of a phantom neck and the design of an RFID reader. The expandable design presented in Chapter 4 has the capability to attach a TRF7960 NFC reader IC (Texas Instruments, Dallas Texas) by SPI for triggering and receiving data from the reader. A “phantom” neck model has been designed to stimulate

the swallowing action by moving a pill through a tube surrounded by tissue simulant that roughly mimics the electrical properties and anatomy of the neck. This test setup will be instrumental in the design and measurement of RFID antennas for both the coil on the exterior of the neck and a coil connected to the RFID tag inside the pill capsule. Future work will be done on the design of the coil which will undergo testing on the phantom neck for feasibility before it is implemented on human subjects.

5.2.3 Algorithm Development

Though a design of a system for collection of data has been presented, the development of algorithms will be a key feature in the use of the device. By making use of the multimodal system, this design is poised to provide better accuracy over systems implementing only one strategy. Work on algorithms for tracheal activities is already underway for the WEAMS system with good results in real-time detection of tracheal activities based on classifications features extracted through audio only[40]. Future work in this area will include the inclusion of other modalities to the audio sensing for higher accuracy with the goal of monitoring pill swallows, food intake, and other tracheal activities. Additionally, making use of the system as a platform for other systems, development of many other algorithms may be possible including gait analysis for predicting falls, and the monitoring of heart activity through PPG. The expandable design of the platform allows algorithm development for many vital signs that may be accessed at the neck.

5.2.4 Smart WEAMS System

The work presented in this thesis focused on the development of a preliminary neck-worn system which acted as a data collection device. The WEAMS prototype

described in chapter 4 requires tethering to a receiver in order for data to be captured and processed. In order to be used as a stand-alone, unit a device must have some independence from a receiver. Independence requires the ability to store data and process it to some extent without the need for an for continuous communication with another device. Such a system has been proposed but is an area of future work for the continuation of the WEAMS project. The next version of the WEAMS device with processing capability, dubbed the Smart WEAMS (SWEAMS) will need to implement a design that is capable of performing digital signal processing in real-time for the automatic classification of swallow data using integrated signal processing for triggering of tracheal activities. The SWEAMS should also have some amount of onboard storage available for holding of data when a smartphone or other base platform is not available.

The proposal for the SWEAMS system includes a higher performing microcontroller such as the ARM Cortex M3 and a low power FPGA for digital signal processing. The ARM processor is a upgrade from the CC2541 MCU currently being used due to its ability to be interfaced with flash storage and having a more modern processor design with higher performance. The inclusion of an FPGA would allow a low power listening device which could process data based on algorithm and wake upon some threshold being met, considerable lowering the power requirements of the system. In addition, the inclusion of internal storage will allow the device to operate without using a wireless link furthering power reduction of the total system.

The SWEAMS device is to include these processing capability and local storage in addition to the sensors available on the WEAMS. A dual mode Bluetooth chip for transmission of both Bluetooth audio and BLE data may also be implemented to send

audio data to a receiver without the need for an additional Bluetooth audio board as is currently implemented. Integration of these components allows for an inclusive design that operates on both audio and other sensor data autonomously.

5.3 Final Words

Though more must be done to develop a neck based wearable platform for detection of health markers, the work here shows feasibility of such a design and lays the groundwork for future work on a novel system that may collect data from a plethora of vital signs from this unique location. With the development of further algorithms, fashion implications, and user interface the design the design may be used in the pursuit of providing a beneficial healthcare aid and personalized health experiences.

APPENDIX A

COLLECTION OF COMMERCIAL WEARABLE DEVICES

Table 4. Commercial Consumer Use Devices

Name	Company	Location	Uses	Sensors
Flex [110]	FitBit	Wrist	Activity Tracking	Inertial (Accelerometer)
Surge [111]	FitBit	Wrist	Activity Tracking, Heart Rate Monitor	Inertial(Gyroscope, Accelerometer), Heart Rate (Optical), GPS, Altimeter, Ambient Light, Compass (Magnetometer),
Charge (HR) [112]	FitBit	Wrist	Activity Tracking , Heart Rate Monitor	Inertial Heart Rate (Optical)
Microsoft- Band [113]	Microsoft	Wrist	Activity Tracking ,(Steps, Run), Calorie Tracking, Heart Rate Monitor, Sleep Tracking, UV Monitor	Heart Rate Monitor (Optical), Galvanic Sensor, Skin Temperature, Microphone, Inertial (Gyro, Accelerometer), Ambient Light
Peak [114]	Basis	Wrist	Activity , Perspiration, Skin Temp, Calorie Tracking, Heart Rate, Sleep Tracking	Inertial (Accelerometer) Temperature Gavanic Sensor Heart Rate (Optical)

Table 2 (continued)

Moov [115]	Moov	Wrist	Activity Tracking	Inertial(Gyroscope, Accelerometer), Compass (Magnetometer)
UP 3 [116]	JawBone	Wrist	Activity Tracking, Sleep Tracking, Heart Rate	Heart Rate (Electrical BioImpedance), Respiration(Electrical BioImpedance), Gavanic Skin Response Inertial (Accelerometer), Ambient Temperature
SENSO-GRAM [117]	Senso-Track	Ear	Activity Tracking, Heart Rate	Inertial(Gyroscope,Accelerometer), Optical PPG
VivoSmart [118]	Garmin	Wrist	Activity Tacking , Sleep Tracking	Inertial, Heart Rate
Misfit Shine [119]	Misfit Wearables	Multiple Locations	Activity Tracking	Accelerometer
iHealth Edge [120]	iHealth	Wrist	Activity Tracking, Sleep Tracking	Accelerometer
LifeBand Touch [121]	LG	Wrist	Activity Tracking	Accelerometer Altimeter
Gear Fit [122]	Samsung	Wrist	Activity Tracking, Heart Rate	Optical PPG, Accelerometer, Gyroscope
BioHarness [123]	Zephyr	Chest Patch	Activity tracking	Inertial Sensors, GPS, ECG
SenseWear [124]	Body-Media	ArmBand	Activity tracking, Energy Expenditure	GSR, Temperature, Accelerometer

Table 5. Commercial Devices Intended for Medical Use

Name	Company	Location	Uses	Sensors	Focus
BodyGuardian [125]	Preventice	Chest	ECG Activity Monitoring Body Position Detection	Electrical ECG	Multi
Proteus Digital Feedback [126]	Proteus	Chest Patch/Multi Placement	ECG Skin Conductance Other capabilities based on sensors possible Pill Ingestion	ECG AFE Impedance Measurement Accelerometer Temperature Galvanic Sensor RFID reader	Multi
HealthPatch [127]	Vital Connect	Chest Patch	Fall Detection EMG Temperature	Accelerometer ECG(Single Lead) Thermistor	Multi
Neuma BioWatch [128]	Neumitra	Wrist	Stress Sensing	Galvanic Skin response Temp Sensor Accelerometer	Multi

Table 3 (continued)

AVIVO (PiiX) [129]	Corventis	Chest Patch	ECG Heart Rate Posture Activity Respiration Bodily Fluid Status Temperature	BioImpedance Sensor (Impedance Plethysmogram ECG	Heart Failure
Metria [130]	Vancive	Arm Patch	Activity Tracking	Inertial Sensors	Activity
PAMSys-ECG [131]	BioSensics	Necklace	Activity Gait Fall Detection	Inertial Sensors	Research Platform
BalanSens [132]	BioSensics	Belt	Balance Detection	Inertial Sensors	Gait
ViSi Mobil [133]	Sotera Wireless	Multiple Placement	ECG Heart Rate SpO2 Blood Pressure	Pulse Oximeter ECG Electrodes(3 position) Pressure Cuff Sensor	Hospital Use

APPENDIX B

RECENT WEARABLE DEVICES FROM RESEARCH INSTITUTIONS

Table 6. New Health Monitors in Research from 2013 Onward

Name/ Citation	Institution	Body Location	Sensors	Uses
[82]	UCLA	Neck	Flex Sensor	Swallow Detection
Gait Assist [134]	Wearable Computing Lab , Zurich Switzerland	Leg	Accelerometer	Parkinson's Gait
SoliBand [10]	NC State/NSF Assist	Arm	PPG	Self- Powering PPG
[11]	University of Alabama	Jaw	Flex, Accelerometer	Chewing Detection
[74]	Chosun University, Korea	Neck	Accelerometer , gyroscope	Fall detection
[135]	Korea Electronics Technology Institute	Chest (Shirt)	ECG	Wireless ECG
[135]	Massachusetts Institute of Technology	Ear	EEG	Wireless Neural Recording

REFERENCES

- [1] R. Suzman and J. Beard, "Global Health and Aging," 2011.
- [2] "The State of Aging and Health in America 2013," 2013. [Online]. Available: http://www.cdc.gov/features/agingandhealth/state_of_aging_and_health_in_america_2013.pdf.
- [3] "Chronic Diseases at a Glance 2009," 2009. [Online]. Available: <http://www.cdc.gov/chronicdisease/resources/publications/AAG/pdf/chronic.pdf>. [Accessed: 01-Nov-2014].
- [4] B. W. Ward, J. S. Schiller, and R. A. Goodman, "Multiple Chronic Conditions Among US Adults: A 2012 Update," *Prev. Chronic Dis.*, vol. 11, p. E62, 2014.
- [5] G. Eysenbach, "What is e-health?," *J. Med. Internet Res.*, vol. 3, no. 2, p. E20.
- [6] X.-F. Teng, Y.-T. Zhang, C. C. Y. Poon, and P. Bonato, "Wearable Medical Systems for p-Health," *Biomed. Eng. IEEE Rev.*, vol. 1, pp. 62–74, 2008.
- [7] Y. M. Choi, T. Olubanjo, A. Farajidavar, and M. Ghovanloo, "Potential barriers in adoption of a medication compliance neckwear by elderly population," in *Engineering in Medicine and Biology Society (EMBC), 2013 35th Annual International Conference of the IEEE*, 2013, pp. 4678–4681.
- [8] R. DeVaul, M. Sung, J. Gips, and a. Pentland, "MITHril 2003: applications and architecture," *Seventh IEEE Int. Symp. Wearable Comput. 2003. Proceedings.*, pp. 4–11, 2003.
- [9] D. Malan and T. Fulford-Jones, "Codeblue: An ad hoc sensor network infrastructure for emergency medical care," ... *Work. wearable ...*, pp. 3–6, 2004.
- [10] J. P. Dieffenderfer, E. Beppler, T. Novak, E. Whitmire, R. Jayakumar, C. Randall, W. Qu, R. Rajagopalan, and A. Bozkurt, "Solar powered wrist worn acquisition system for continuous photoplethysmogram monitoring," in *Engineering in Medicine and Biology Society (EMBC), 2014 36th Annual International Conference of the IEEE*, 2014, pp. 3142–3145.

- [11] J. M. Fontana, M. Farooq, and E. Sazonov, "Automatic Ingestion Monitor: A Novel Wearable Device for Monitoring of Ingestive Behavior," in *Biomedical Engineering, IEEE Transactions on*, 2014, vol. 61, no. 6, pp. 1772–1779.
- [12] S. Patel, H. Park, P. Bonato, L. Chan, and M. Rodgers, "A review of wearable sensors and systems with application in rehabilitation.," *J. Neuroeng. Rehabil.*, vol. 9, no. 1, p. 21, Jan. 2012.
- [13] "SHIMMER." [Online]. Available: <http://www.shimmersensing.com/research-and-education/equipment/wireless-sensor-platform-specification/>.
- [14] W. Cui, L. E. Ostrander, and B. Y. Lee, "In vivo reflectance of blood and tissue as a function of light wavelength," *Biomed. Eng. IEEE Trans.*, vol. 37, no. 6, pp. 632–639, 1990.
- [15] J. Lee, K. Matsumura, K.-I. Yamakoshi, P. Rolfe, S. Tanaka, and T. Yamakoshi, "Comparison between red, green and blue light reflection photoplethysmography for heart rate monitoring during motion," in *Engineering in Medicine and Biology Society (EMBC), 2013 35th Annual International Conference of the IEEE*, 2013, pp. 1724–1727.
- [16] J. G. Webster, Ed., *Design of Pulse Oximeters*, 1st ed. CRC Press, 1997.
- [17] Y. Mendelson and C. Pujary, "Measurement site and photodetector size considerations in optimizing power consumption of a wearable reflectance pulse oximeter," in *Engineering in Medicine and Biology Society, 2003. Proceedings of the 25th Annual International Conference of the IEEE*, 2003, vol. 4, pp. 3016–3019 Vol.4.
- [18] Y. Mendelson, J. C. Kent, B. L. Yocum, and M. J. Birle, "Design and evaluation of a new reflectance pulse oximeter sensor.," *Med. Instrum.*, vol. 22, no. 4, pp. 167–173, Aug. 1988.
- [19] W. S. Johnston and Y. Mendelson, "Extracting breathing rate information from a wearable reflectance pulse oximeter sensor," *Engineering in Medicine and Biology Society, 2004. IEMBS '04. 26th Annual International Conference of the IEEE*, vol. 2, pp. 5388–5391, 2004.
- [20] R. Matthews, N. J. McDonald, P. Hervieux, P. J. Turner, and M. A. Steindorf, "A Wearable Physiological Sensor Suite for Unobtrusive Monitoring of Physiological and Cognitive State," in *Engineering in Medicine and Biology Society, 2007. EMBS 2007. 29th Annual International Conference of the IEEE*, 2007, pp. 5276–5281.

- [21] Y. M. Chi and G. Cauwenberghs, "Wireless Non-contact EEG/ECG Electrodes for Body Sensor Networks," *2010 Int. Conf. Body Sens. Networks*, pp. 297–301, Jun. 2010.
- [22] A. M. Chan, N. Selvaraj, N. Ferdosi, and R. Narasimhan, "Wireless patch sensor for remote monitoring of heart rate, respiration, activity, and falls,," in *Conference proceedings : ... Annual International Conference of the IEEE Engineering in Medicine and Biology Society. IEEE Engineering in Medicine and Biology Society. Annual Conference*, 2013, vol. 2013, pp. 6115–8.
- [23] V. Mihajlovic, B. Grundlehner, R. Vullers, and J. Penders, "Wearable, Wireless EEG Solutions in Daily Life Applications: What are we missing?," *Biomed. Heal. Informatics, IEEE J.*, vol. PP, no. 99, p. 1, 2014.
- [24] A. Allahverdy, A. M. Nasrabadi, and M. R. Mohammadi, "Detecting ADHD children using symbolic dynamic of nonlinear features of EEG," in *Electrical Engineering (ICEE), 2011 19th Iranian Conference on*, 2011, pp. 1–4.
- [25] S. Iampetch, Y. Punsawad, and Y. Wongsawat, "EEG-based mental fatigue prediction for driving application," in *Biomedical Engineering International Conference (BMEiCON), 2012*, 2012, pp. 1–5.
- [26] S. Grimnes and Ø. G. Martinsen, *Bioimpedance and Bioelectricity Basics*. Elsevier, 2000, pp. 1–2.
- [27] C. Cornelius and C. N. Gutierrez, "A Survey of Biometrics for Wearable Devices Contributors," *Intel Technol. J.*, vol. 18, no. 4, pp. 46–64, 2014.
- [28] I. Cuba-Gyllensten, P. Gastelurrutia, J. Riistama, R. Aarts, J. Nuñez, J. Lupon, and A. Bayes-Genis, "A novel wearable vest for tracking pulmonary congestion in acutely decompensated heart failure," *Int. J. Cardiol.*, vol. 177, no. 1, pp. 199–201, Nov. 2014.
- [29] J. Ferreira, F. Seoane, and K. Lindecrantz, "Portable bioimpedance monitor evaluation for continuous impedance measurements. Towards wearable plethysmography applications," *Eng. Med. Biol. Soc. (EMBC), 2013 35th Annu. Int. Conf. IEEE*, pp. 559–562, 2013.
- [30] P. Batra and R. Kapoor, "A Novel Method for Heart Rate Measurement Using Bioimpedance," *Advances in Recent Technologies in Communication and Computing (ARTCom), 2010 International Conference on*. pp. 443–445, 2010.
- [31] G. Medrano, L. Beckmann, N. Zimmerman, T. Grundmann, T. Gries, and S. Leonhardt, "Bioimpedance spectroscopy with textile electrodes for a continuous monitoring application," in *4th International Workshop on Wearable and Implantable Body Sensor Networks*, 2007, pp. 23–28.

- [32] G. Wu, G. Liu, and M. Hao, "The Analysis of Emotion Recognition from GSR Based on PSO," in *Intelligence Information Processing and Trusted Computing (IPTC), 2010 International Symposium on*, 2010, pp. 360–363.
- [33] M.-Z. Poh, T. Loddenkemper, N. C. Swenson, S. Goyal, J. R. Madsen, and R. W. Picard, "Continuous monitoring of electrodermal activity during epileptic seizures using a wearable sensor," in *Engineering in Medicine and Biology Society (EMBC), 2010 Annual International Conference of the IEEE*, 2010, pp. 4415–4418.
- [34] K. Khalfallah, H. Ayoub, J. H. Calvet, X. Neveu, P. Brunswick, S. Griveau, V. Lair, M. Cassir, and F. Bedioui, "Noninvasive Galvanic Skin Sensor for Early Diagnosis of Sudomotor Dysfunction: Application to Diabetes," *Sensors Journal, IEEE*, vol. 12, no. 3, pp. 456–463, 2012.
- [35] D. Oletic and V. Bilas, "Wireless sensor node for respiratory sounds monitoring," in *Instrumentation and Measurement Technology Conference (I2MTC), 2012 IEEE International*, 2012, pp. 28–32.
- [36] W. Ser, Z.-L. Yu, J. Zhang, and J. Yu, "Wearable system design with wheeze signal detection," in *Medical Devices and Biosensors, 2008. ISSS-MDBS 2008. 5th International Summer School and Symposium on*, 2008, pp. 260–263.
- [37] K. Yatani and K. N. Truong, "BodyScope: A Wearable Acoustic Sensor for Activity Recognition," in *ACM International Conference on Ubiquitous Computing*, 2012.
- [38] K. Hung, B. L. Luk, W. H. Choy, B. Tai, and S. K. Tso, "Multifunction stethoscope for telemedicine," *Computer Architectures for Machine Perception, 2003 IEEE International Workshop on*, pp. 87–89, 2004.
- [39] T. Rahman, A. Adams, M. Zhang, E. Cherry, B. Zhou, H. Peng, and T. Choudhury, "BodyBeat: a mobile system for sensing non-speech body sounds," in *MobiSys '14 Proceedings of the 12th annual international conference on Mobile systems, applications, and services*, 2014, pp. 2–13.
- [40] T. Olubango and M. Ghovanloo, "Tracheal activity recognition based on acoustic signals," *Engineering in Medicine and Biology Society (EMBC), 2014 36th Annual International Conference of the IEEE*, pp. 1436–1439, 2014.
- [41] E. S. Sazonov, O. Makeyev, S. Schuckers, P. Lopez-Meyer, E. L. Melanson, and M. R. Neuman, "Automatic Detection of Swallowing Events by Acoustical Means for Applications of Monitoring of Ingestive Behavior," *Biomed. Eng. IEEE Trans.*, vol. 57, no. 3, pp. 626–633, 2010.

- [42] S. Passler and W. Fischer, "Food Intake Activity Detection Using a Wearable Microphone System," in *Intelligent Environments (IE), 2011 7th International Conference on*, 2011, pp. 298–301.
- [43] A. Lanata, E. P. Scilingo, and D. De Rossi, "A Multimodal Transducer for Cardiopulmonary Activity Monitoring in Emergency," *Inf. Technol. Biomed. IEEE Trans.*, vol. 14, no. 3, pp. 817–825, 2010.
- [44] K. Hung, C. C. Lee, W. M. Chan, S.-O. Choy, and P. Kwok, "Development of a wearable system integrated with novel biomedical sensors for ubiquitous healthcare," in *Engineering in Medicine and Biology Society (EMBC), 2012 Annual International Conference of the IEEE*, 2012, pp. 5802–5805.
- [45] K. Budidha and P. a Kyriacou, "The human ear canal: investigation of its suitability for monitoring photoplethysmographs and arterial oxygen saturation.," *Physiol. Meas.*, vol. 35, no. 2, pp. 111–28, Feb. 2014.
- [46] H. Kalantarian, S. I. Lee, A. Mishra, H. Ghasemzadeh, J. Liu, and M. Sarrafzadeh, "Multimodal Energy Expenditure Calculation for Pervasive Health : A Data Fusion Model using Wearable Sensors," in *5t International Workshop on Smart Environments and Abient Intelligence*, 2013, no. March, pp. 676–681.
- [47] M. Caldara, C. Colleoni, E. Guido, G. Rosace, V. Re, and A. Vitali, "A wearable sensor platform to monitor sweat pH and skin temperature," in *2013 IEEE International Conference on Body Sensor Networks*, 2013, pp. 1–6.
- [48] B. Liu, Y. Zhang, and Z. Liu, "Wearable monitoring system with multiple physiological parameters," *Medical Devices and Biosensors, 2008. ISSS-MDBS 2008. 5th International Summer School and Symposium on*. pp. 268–271, 2008.
- [49] Y. Varatharajah, N. Karunathilaka, M. Rismi, S. Kotinkaduwa, and D. Dias, "Body area sensor network for evaluating fitness exercise," in *Wireless and Mobile Networking Conference (WMNC), 2013 6th Joint IFIP*, 2013, pp. 1–8.
- [50] E. S. Sazonov and J. M. Fontana, "A Sensor System for Automatic Detection of Food Intake Through Non-Invasive Monitoring of Chewing," *Sensors Journal, IEEE*, vol. 12, no. 5. pp. 1340–1348, 2012.
- [51] D. D. He, E. S. Winokur, and C. G. Sodini, "A continuous, wearable, and wireless heart monitor using head ballistocardiogram (BCG) and head electrocardiogram (ECG)," in *Engineering in Medicine and Biology Society, EMBC, 2011 Annual International Conference of the IEEE*, 2011, pp. 4729–4732.
- [52] P. Castiglioni, A. Faini, G. Parati, and M. Di Rienzo, "Wearable Seismocardiography," *Engineering in Medicine and Biology Society, 2007. EMBS 2007. 29th Annual International Conference of the IEEE*. pp. 3954–3957, 2007.

- [53] D. H. Phan, S. Bonnet, R. Guillemaud, E. Castelli, and N. Y. Pham Thi, "Estimation of respiratory waveform and heart rate using an accelerometer," in *Engineering in Medicine and Biology Society, 2008. EMBS 2008. 30th Annual International Conference of the IEEE*, 2008, pp. 4916–4919.
- [54] M. Kangas, A. Konttila, P. Lindgren, I. Winblad, and T. Jämsä, "Comparison of low-complexity fall detection algorithms for body attached accelerometers.," *Gait Posture*, vol. 28, no. 2, pp. 285–91, Aug. 2008.
- [55] W. Ludwig, K.-H. Wolf, C. Duwenkamp, N. Gusew, N. Hellrung, M. Marschollek, M. Wagner, and R. Haux, "Health-enabling technologies for the elderly--an overview of services based on a literature review.," *Comput. Methods Programs Biomed.*, vol. 106, no. 2, pp. 70–8, May 2012.
- [56] S. Morales-Vidal and S. Ruland, "Telemedicine in stroke care and rehabilitation.," *Top. Stroke Rehabil.*, vol. 20, no. 2, pp. 101–7, 2013.
- [57] N. Sazonova, R. Browning, E. Melanson, and E. Sazonov, "Posture and activity recognition and energy expenditure prediction in a wearable platform," in *Engineering in Medicine and Biology Society (EMBC), 2014 36th Annual International Conference of the IEEE*, 2014, pp. 4163–4167.
- [58] L. Atallah, B. Lo, R. King, and G.-Z. Yang, "Sensor Positioning for Activity Recognition Using Wearable Accelerometers," *Biomed. Circuits Syst. IEEE Trans.*, vol. 5, no. 4, pp. 320–329, 2011.
- [59] J. Chakaravarthy, "UP3: THE FIRST TRULY WEARABLE HEART RATE MONITOR," 2014. [Online]. Available: <https://jawbone.com/blog/up3-wearable-heart-rate-monitor/>. [Accessed: 16-Nov-2014].
- [60] S. S. Thomas, V. Nathan, C. Zong, E. Akinbola, A. L. P. Aroul, L. Philipose, K. Soundarapandian, X. Shi, and R. Jafari, "BioWatch - A wrist watch based signal acquisition system for physiological signals including blood pressure," in *Engineering in Medicine and Biology Society (EMBC), 2014 36th Annual International Conference of the IEEE*, 2014, pp. 2286–2289.
- [61] M.-Z. Poh, N. C. Swenson, and R. W. Picard, "A Wearable Sensor for Unobtrusive, Long-Term Assessment of Electrodermal Activity," *Biomed. Eng. IEEE Trans.*, vol. 57, no. 5, pp. 1243–1252, 2010.
- [62] J. M. Kang, T. Yoo, and H. C. Kim, "A Wrist-Worn Integrated Health Monitoring Instrument with a Tele-Reporting Device for Telemedicine and Telecare," *Instrum. Meas. IEEE Trans.*, vol. 55, no. 5, pp. 1655–1661, 2006.

- [63] “Proteus Digital Health.” [Online]. Available: <http://www.proteus.com/technology/developer-network/>. [Accessed: 11-Apr-2014].
- [64] M. Peng, T. Wang, G. Hu, and H. Zhang, “A wearable heart rate belt for ambulant ECG monitoring,” in *e-Health Networking, Applications and Services (Healthcom), 2012 IEEE 14th International Conference on*, 2012, pp. 371–374.
- [65] B. R. Spiegel, M. Kaneshiro, M. Russell, A. Lin, A. Patel, V. Tashjian, V. Zegarski, D. Singh, S. Cohen, M. Reid, C. Whitman, J. Talley, B. Martinez, and W. Kaiser, “Validation of an Acoustic Gastrointestinal Surveillance Biosensor for Postoperative Ileus,” *J. Gastrointest. Surg.*, vol. 18, no. 10, pp. 1795–1803, 2014.
- [66] M. Chan, D. Estève, J.-Y. Fourniols, C. Escriba, and E. Campo, “Smart wearable systems: current status and future challenges,” *Artif. Intell. Med.*, vol. 56, no. 3, pp. 137–56, Nov. 2012.
- [67] Y. Meng and H.-C. Kim, “Wearable Systems and Applications for Healthcare,” in *Computers, Networks, Systems and Industrial Engineering (CNSI), 2011 First ACIS/JNU International Conference on*, 2011, pp. 325–330.
- [68] A. Lymberis and A. Dittmar, “Advanced Wearable Health Systems and Applications - Research and Development Efforts in the European Union,” *Eng. Med. Biol. Mag. IEEE*, vol. 26, no. 3, pp. 29–33, 2007.
- [69] E. Parikesit, T. L. R. Mengko, and H. Zakaria, “Wearable gait measurement system based on accelerometer and pressure sensor,” in *Instrumentation, Communications, Information Technology, and Biomedical Engineering (ICICI-BME), 2011 2nd International Conference on*, 2011, pp. 395–398.
- [70] E. S. Sazonov, G. Fulk, N. Sazonova, and S. Schuckers, “Automatic recognition of postures and activities in stroke patients,” *Conf. Proc. ... Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. IEEE Eng. Med. Biol. Soc. Annu. Conf.*, vol. 2009, pp. 2200–2203, 2009.
- [71] B. M. Trindade, Y. Ono, E. D. Lemaire, and I. AlMohimeed, “Development of a wearable ultrasonic sensor and method for continuous monitoring of mechanical properties of plantar soft tissue for diabetic patients,” in *Ultrasonics Symposium (IUS), 2014 IEEE International*, 2014, pp. 2112–2115.
- [72] J. Healey, “GSR Sock: A New e-Textile Sensor Prototype,” *Wearable Computers (ISWC), 2011 15th Annual International Symposium on*, pp. 113–114, 2011.
- [73] Y.-P. Hsu and D. J. Young, “Skin-surface-coupled personal health monitoring system,” *SENSORS, 2013 IEEE*, pp. 1–4, 2013.

- [74] W.-S. Baek, D.-M. Kim, F. Bashir, and J.-Y. Pyun, "Real life applicable fall detection system based on wireless body area network," in *Consumer Communications and Networking Conference (CCNC), 2013 IEEE*, 2013, pp. 62–67.
- [75] A. Mizuno, H. Okumura, and M. Matsumura, "Development of neckband mounted active bio-electrodes for non-restraint lead method of ECG R wave," in *4th European Conference of the International Federation for Medical and Biological Engineering SE - 330*, vol. 22, J. Vander Sloten, P. Verdonck, M. Nyssen, and J. Haueisen, Eds. Springer Berlin Heidelberg, 2009, pp. 1394–1397.
- [76] A. Jubran, "Pulse oximetry," *Crit. Care*, no. 3, pp. 11–17, 1999.
- [77] D. Da He, E. S. Winokur, T. Heldt, and C. G. Sodini, "The ear as a location for wearable vital signs monitoring,," in *Conference proceedings : ... Annual International Conference of the IEEE Engineering in Medicine and Biology Society. IEEE Engineering in Medicine and Biology Society. Annual Conference*, 2010, vol. 2010, pp. 6389–6392.
- [78] P. Kidmose, D. Looney, L. Jochumsen, and D. P. Mandic, "Ear-EEG from generic earpieces: A feasibility study," *Engineering in Medicine and Biology Society (EMBC), 2013 35th Annual International Conference of the IEEE*. pp. 543–546, 2013.
- [79] T. Kusuhara, T. Nakamura, K. Mori, and Y. Yamamoto, "The adapting method of impedance pharyngography (IPG) to nodding swallowing," *Biomedical Engineering, 2003. IEEE EMBS Asian-Pacific Conference on*. pp. 286–287, 2003.
- [80] T. Nakamura, Y. Yamamoto, and H. Tsugawa, "Measurement system for swallowing based on impedance pharyngography and swallowing sound," *Instrumentation and Measurement Technology Conference, 2000. IMTC 2000. Proceedings of the 17th IEEE*, vol. 1. pp. 191–194 vol.1, 2000.
- [81] O. Amft and G. Troster, "Methods for Detection and Classification of Normal Swallowing from Muscle Activation and Sound," *Pervasive Health Conference and Workshops, 2006*. pp. 1–10, 2006.
- [82] H. Kalantarian, N. Alshurafa, and M. Sarrafzadeh, "A Wearable Nutrition Monitoring System," in *2014 11th International Conference on Wearable and Implantable Body Sensor Networks*, 2014, pp. 75–80.
- [83] H. Cao, X. Liang, I. Balasingham, and V. C. M. Leung, "Performance Analysis of ZigBee Technology for Wireless," in *Ad Hoc Networks: First International Conference on*, 2009, pp. 747–761.

- [84] X. Hu, J. Wang, Q. Yu, W. Liu, and J. Qin, "A Wireless Sensor Network Based on ZigBee for Telemedicine Monitoring System," in *Bioinformatics and Biomedical Engineering, 2008. ICBBE 2008. The 2nd International Conference on*, 2008, pp. 1367–1370.
- [85] P. Smith, "Comparing Low-Power Wireless Technologies," 2011.
- [86] "Specification of the Bluetooth System," vol. 0, no. December, 2013.
- [87] F. Klaus, *RFID handbook*, 3rd ed. 2010.
- [88] L. Yang, R. Vyas, A. Rida, J. Pan, and M. M. Tentzeris, "Wearable RFID-enabled sensor nodes for biomedical applications," *Electronic Components and Technology Conference, 2008. ECTC 2008. 58th*. pp. 2156–2159, 2008.
- [89] H. Rajagopalan and Y. Rahmat-Samii, "Platform tolerant and capsule-pill RFID antenna designs: An overview of recent developments at UCLA," *Antenna Technology (iWAT), 2012 IEEE International Workshop on*. pp. 144–147, 2012.
- [90] "SensorTag." [Online]. Available: http://www.ti.com/ww/en/wireless_connectivity/sensortag/index.shtml?DCMP=senortag&HQS=sensortag-bn. [Accessed: 20-Nov-2014].
- [91] "IASUS CONCEPTS - NT3 Throat Mic System." [Online]. Available: <http://www.iasus-concepts.com/product?pn=NT3-Throat-Mic-System><http://www.iasus-concepts.com/product?pn=NT3-Throat-Mic-System>. [Accessed: 20-Nov-2014].
- [92] "Piezo Sensor - DT Series Lead Attachments." [Online]. Available: http://www.meas-spec.com/product/t_product.aspx?id=2480. [Accessed: 20-Nov-2014].
- [93] A. Andersen, "Small Size 2.4 GHz PCB antenna." [Online]. Available: <http://www.ti.com/lit/an/swra117d/swra117d.pdf>.
- [94] "Invesense - MPU9250." [Online]. Available: <http://www.invensense.com/mems/gyro/mpu9250.html>. [Accessed: 20-Nov-2014].
- [95] "AFE4400." [Online]. Available: <http://www.ti.com/product/AFE4400/datasheet>. [Accessed: 19-Nov-2014].
- [96] "AFE4490." [Online]. Available: AFE4400. (n.d.). Retrieved November 19, 2014, from <http://www.ti.com/product/AFE4490/datasheet>. [Accessed: 19-Nov-2014].
- [97] "DCM05." [Online]. Available: http://www.apmkr.com/bio-device/DCM05_datasheet.pdf. [Accessed: 16-Nov-2014].

- [98] “INA333.” [Online]. Available: <http://www.ti.com/product/ina333>. [Accessed: 19-Nov-2014].
- [99] “LTC4054-4.2 - Standalone Linear Li-Ion Battery Charger with Thermal Regulation in ThinSOT.” [Online]. Available: <http://www.linear.com/product/LTC4054-4.2>. [Accessed: 20-Nov-2014].
- [100] A. Chattopadhyay, “Current Savings in CC254x Using the TPS62730,” Dallas Texas.
- [101] “TPS62730.” [Online]. Available: <http://www.ti.com/product/tps62730>. [Accessed: 20-Nov-2014].
- [102] “TPS7A49xx.” [Online]. Available: <http://www.ti.com/lit/ds/symlink/tps7a4901.pdf>. [Accessed: 18-Nov-2014].
- [103] “HomeLab.” [Online]. Available: <http://homelab.gtri.gatech.edu/welcome/>. [Accessed: 16-Nov-2014].
- [104] E. T. Gonzalez, A. M. Jones, L. R. Harley, D. Burnham, Y. M. Choi, W. B. Fain, and M. Ghovanloo, “Older Adults’ Perceptions of a Neckwear Health Technology,” in *Proceedings of the Human Factors and Ergonomics Society Annual Meeting*, 2014, vol. 58, no. 1, pp. 1815–1819.
- [105] “Google Fit.” [Online]. Available: <https://developers.google.com/fit/>. [Accessed: 19-Nov-2014].
- [106] “Health.” [Online]. Available: <https://www.apple.com/ios/whats-new/health/>. [Accessed: 19-Nov-2014].
- [107] Cadence, “HiFi Audio, Voice, and Speech.” [Online]. Available: <http://ip.cadence.com/ipportfolio/tensilica-ip/audio>. [Accessed: 15-Dec-2014].
- [108] J. Yoshida, “Wearable SoC: Let DSP Do ‘Always Listening’ Chores.” [Online]. Available: http://www.eetimes.com/document.asp?doc_id=1322279. [Accessed: 22-Dec-2014].
- [109] D. Du and K. M. Odame, “A Bandwidth-Adaptive Preamplifier,” *Solid-State Circuits, IEEE J.*, vol. 48, no. 9, pp. 2142–2153, 2013.
- [110] “FitBit Flex.” [Online]. Available: <http://www.fitbit.com/flex>. [Accessed: 20-Nov-2014].
- [111] “FitBit Surge.” [Online]. Available: <https://www.fitbit.com/surge>. [Accessed: 20-Nov-2014].

- [112] "FitBit Charge." [Online]. Available: <https://www.fitbit.com/charge>. [Accessed: 20-Nov-2014].
- [113] "Microsoft Band." [Online]. Available: <http://www.microsoft.com/microsoft-band/en-us>. [Accessed: 20-Nov-2014].
- [114] "Basis Peak." [Online]. Available: <http://www.mybasis.com/>. [Accessed: 20-Nov-2014].
- [115] "Moov." [Online]. Available: <http://preorder.now.moov.cc/>. [Accessed: 20-Nov-2014].
- [116] "Jawbone Up 3 - Wearable HeartRate Monitor." [Online]. Available: <https://jawbone.com/blog/up3-wearable-heart-rate-monitor/>.
- [117] "SENSOTRACK." [Online]. Available: <http://www.sensotrack.com/>. [Accessed: 20-Nov-2014].
- [118] "Vivosmart." [Online]. Available: <http://sites.garmin.com/en-US/vivo/vivosmart/>. [Accessed: 20-Nov-2014].
- [119] "MISFIT." [Online]. Available: <http://misfit.com/>. [Accessed: 20-Nov-2014].
- [120] "iHealth Edge." [Online]. Available: www.ihealthlabs.com/fitness-devices/ihealth-edge/. [Accessed: 20-Nov-2014].
- [121] "Fitness & Activity Trackers - LG Electronics." [Online]. Available: www.lg.com/us/fitness-activity-trackers. [Accessed: 20-Nov-2014].
- [122] "Samsung Gear Fit TM." [Online]. Available: http://www.samsung.com/global/microsite/gear/gearfit_features.html. [Accessed: 20-Nov-2014].
- [123] "Zephyr BioHarness 3 - Wireless Professional Heart Rate Monitor." [Online]. Available: <http://www.zephyranywherestore.com/BioHarness-Wireless-Professional-Physiological-Bluetooth/dp/B009ZUWWG6>. [Accessed: 20-Nov-2014].
- [124] "SenseWear." [Online]. Available: <http://sensewear.bodymedia.com/>. [Accessed: 20-Nov-2014].
- [125] "BodyGuardian - Preventice." [Online]. Available: www.preventice.com/products/bodyguardian. [Accessed: 20-Nov-2014].

- [126] “Proteus Wearable Sensing for Developers.” [Online]. Available: <http://www.proteusdigitalhealth.com/assets/Proteus-Wearable-Sensing-for-Developers.pdf>. [Accessed: 20-Nov-2014].
- [127] “VitalConnect.” [Online]. Available: <http://www.vitalconnect.com/overview>. [Accessed: 20-Nov-2014].
- [128] “Nuema BioWatch.” [Online]. Available: <https://www.neumitra.com/>. [Accessed: 20-Nov-2014].
- [129] “Avivo MPM- How it works.” [Online]. Available: <http://www.corventis.com/products/avivo-mpm/how-it-works/>. [Accessed: 20-Nov-2014].
- [130] “Metria™ Informed Health.” [Online]. Available: <http://www.averydennison.com/en/home/technologies/creative-showcase/metria-wearable-sensor.html>. [Accessed: 20-Nov-2014].
- [131] “Pamsys.” [Online]. Available: www.biosensics.com/pamsys-overview/. [Accessed: 20-Nov-2014].
- [132] “BalanSens.” [Online]. Available: <http://www.biosensics.com/balansens-overview/>. [Accessed: 20-Nov-2014].
- [133] “ViSi Mobile.” [Online]. Available: <http://www.visimobile.com/visi-product-info/technical/>. [Accessed: 20-Nov-2014].
- [134] S. Mazilu, U. Blanke, M. Hardegger, G. Troster, E. Gazit, M. Dorfman, and J. M. Hausdorff, “GaitAssist: A wearable assistant for gait training and rehabilitation in Parkinson’s disease,” in *Pervasive Computing and Communications Workshops (PERCOM Workshops), 2014 IEEE International Conference on*, 2014, pp. 135–137.
- [135] Y. H. Kim, K.-J. Jang, S. Lee, C. W. Park, and H. Y. Youn, “A robust wearable health monitoring system based on WSN,” in *Consumer Communications and Networking Conference (CCNC), 2013 IEEE*, 2013, pp. 288–293.